The Propulsion Dynamics of Human Locomotion

A Thesis

Submitted to the Faculty

of

Drexel University

by

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in partial fulfillment of the

requirements for the degree

of

Doctor of Philosophy

May 2009

Dedications

To my family for their unconditional love and support.

Acknowledgments

I deeply thank my advisors Dr. Tucker and Dr. Seliktar for their continuous support, guidance and friendship.

I thank the staff at Shriners Hospital-Philadelphia motion lab and research department for their assistance and support during my time there.

Finally, my gratitude to the children who participated in this study and their families for donating their time.

Table of Contents

List of Tables

List of Tables (continued)

List of Tables (continued)

List of Figures

List of Figures (continued)

List of Figures (continued)

List of Figures (continued)

Abstract

The Propulsion Dynamics of Human Locomotion

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Gait analysis quantifies biomechanical aspects of an individual's gait. It can be used to quantify abnormalities in gait, identify underlying biomechanical mechanisms and track changes in gait biomechanics over time or in response to surgical or clinical interventions. In clinical gait analysis, kinematic and kinetic data are captured and used for clinical decision making; however, clinicians rely primarily on the kinematic information. Further, gait analyses are commonly performed in the confines of a closed clinical laboratory, and only on level surfaces with limited walking lengths. Another often overlooked component of gait analysis is the quantification of energy expenditure (EE), which is directly impacted by the interplay between kinetics and mechanical energy changes.

The long-term goal of this research is to develop data collection and analytic methods to combine kinetic and kinematic, better quantify the relationship between kinetic and EE data and support means for data collection outside of traditional laboratories. In addition, application to pathological gait, specifically that of children with Cerebral Palsy (CP), was preliminarily assessed. The primary short term goal of this work was to develop analytic means to determine the acceleration of the center of mass (COM) and ground reaction forces (GRFs) during gait that relate to the consistency and stability of the gait pattern, as well as to energy expenditure. Application of a consistency

test based on the force time integrals of the anterior-posterior (A-P) component of the GRF defined the boundaries of a consistent gait both in typically developed (TD) children and children with CP. GRFs characteristics were examined both in TD children and children with CP and the results demonstrated that GRFs characteristics differ between these populations and are capable of detecting gait abnormalities. Good correlations between the peak braking force and EE were found for both populations. An analytical method, consisting of two accelerometers placed on the back, was developed for derivation of GRFs characteristics from the COM's acceleration. The results showed good to excellent correlations to GRF data when several characteristics of the GRFs were compared. The results support the continuation of this work towards the achievement of the long term goal.

Chapter 1: Introduction

Gait analysis is a process that quantifies biomechanical aspects of an individual's gait. Gait analysis can be used to quantify abnormalities in gait, identify underlying biomechanical mechanisms and track changes in gait biomechanics over time or in response to surgical or clinical interventions. Gait analysis most commonly relies on inverse-dynamic modeling of the body as a set of linked rigid segments and anthropometric measures. Kinematic and kinetic data are captured through instrumentation such as camera based motion capture system and force plates. Typically, quantitative gait data are acquired by having an individual walk along a smooth straight pathway, approximately 20 feet long, within a laboratory setting. The data collected provide quantification of the movement of body segment, both translations and joint rotations, also known as kinematic measures. In addition, with the use of force plates, kinetic data, including ground reaction forces, can be obtained. The combination of kinematic, kinetic and anthropometric data can be used in inverse dynamic analysis to calculate net joint moments and powers (Davis 2004). These are typically reported in clinical gait analysis and used for tracking changes in gait and for clinical decision making.

However, clinicians often rely primarily on the kinematic information provided, with limited use of the kinetic data collected during gait analysis. Further, the majority of quantitative gait analyses are performed in the confines of a closed space within a clinical setting only on level surfaces over limited walking lengths and such factors might impact the gait of the subject being tested. Another, often overlooked, component of gait analysis is the quantification of energy expenditure, in particular the interplay between kinetic measures and mechanical energy changes. Current data collection and analysis methods do not capitalize on the inter-connectedness of the various gait analysis data and an analytical model that can bridge and better utilize the available information is needed.

1.1. Specific aims

The long-term goal of this research was to develop data collection and analytic methods to combine kinetic and kinematic data that reflect the interplay between kinetic and energy expenditure (EE) data in gait and that provide the beginning means to support data collection outside of traditional laboratories that rely primarily on force plates to obtain the kinetic data. In addition, application of these methods to pathological gait, specifically that of children with Cerebral Palsy (CP), was preliminarily assessed. In order to achieve this long term goal, the primary short term goals reflected in this current work were to develop analytic means to determine the acceleration of the center of mass (COM) and ground reaction forces during gait that relate to the consistency and stability of the gait pattern, as well as the energy components underlying the gait pattern.
The specific aims of the research were:

1. To validate a hypothesis that gait is a cyclic phenomena through analyses of the impulse of the ground reaction forces.
2. To define and assess parameters of consistency and stability in gait of individuals with typical development and of children with CP. The hypothesis was that

consistent and stable gait can be detected through analysis of ground reaction forces and therefore can track changes and quantify gait deviations.

3. To develop a biomechanical model that can derive ground reaction force characteristics from the acceleration of a calculated COM. The hypothesis is that ground reaction forces are proportional to changes in the body's COM acceleration.

4. To validate the model by comparing model prediction based on acceleration data with measures of ground reaction forces. We hypothesize that the assumed COM's acceleration and ground reaction forces will be highly correlated.

5. To identify the relationship between the derived GRFs, the work being exerted by the body COM, and the mechanical energy expended during ambulation.

Chapter 2: Background

2.1. Measurement of ground reaction forces

Collection and analysis of ground reaction force (GRF) data is done with the use of force plates within a motion analysis laboratory (Figures 2.1 and 2.2).

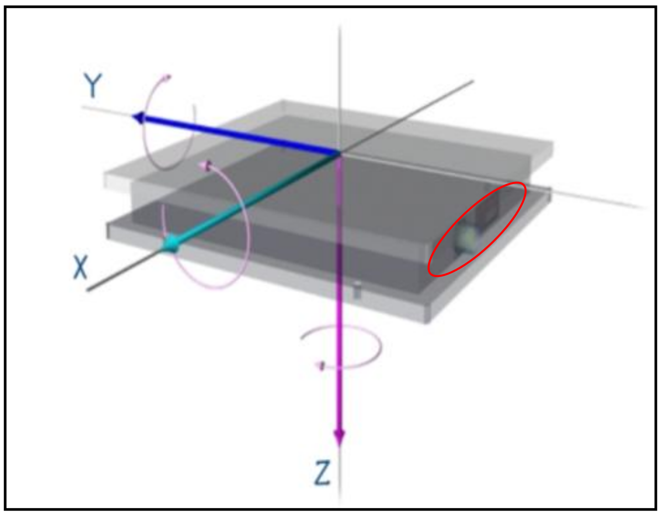

Figure 2.1: Schematic of an AMTI force plate and its coordinate system[50].

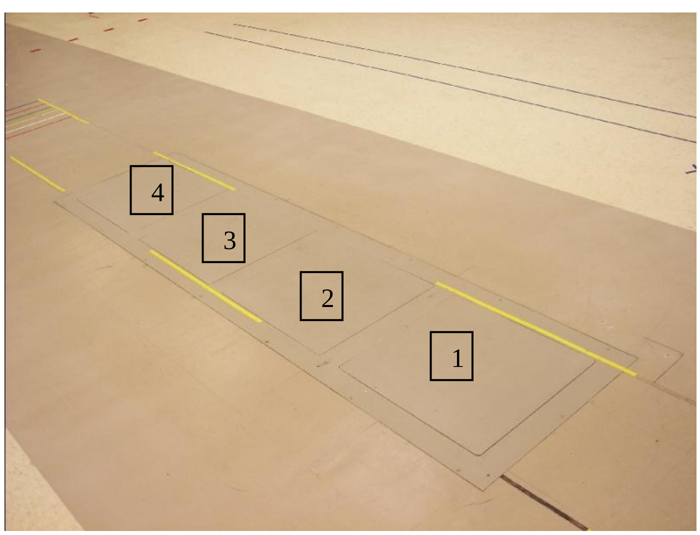

Figure 2.2: Configuration of the motion analysis laboratory with four force plates embedded in the walkway.

As illustrated in Figure 2.1 the force plate is a rectangular plate with a defined orientation of a right hand coordinate system in which the positive z axis is oriented downward. The positive y axis is oriented away from the connector (circled in red) and the positive x axis is oriented to the left when facing in a positive y direction. The force platform measures the forces and moments acting on the top surface of the platform in three components acting along the three coordinate system axes. The AMTI force plate uses strain gauges mounted on precision metal sensing elements located within the platform to perform the force and moment measurement. The strain gauges are electrically wired in full four arm bridge arrangements to provide thermal stability and to isolate the strains caused by forces applied in several directions. In order to function, a strain gauge bridge requires a source of stable excitation voltage. When this voltage is applied across two terminals of the bridge the alternate two terminals of the bridge will be balanced and no signal will be present on those terminals. When a load is applied to the sensing element small mechanical strains will subtly change the resistance of the bridge arms and the bridge will become unbalanced. When this occurs a very small electrical signal will be observed across the bridge. The output signals from the strain gauge bridges must be amplified in order to produce signals of sufficient strength to be useful. Typically an amplifier is used to produce a usable output signal. Once the raw output signals have been amplified they may be digitized using an analog to digital converter. The digitized signals represent values which are proportional to the loads applied to the transducer providing the forces applied to the plate (Nigg, 1999).

2.2. Ground reaction forces during gait

As mentioned previously, gait analysis is a process that quantifies biomechanical aspects of an individual’s gait. When dealing with human gait it is customary to do so in terms of the “gait cycle”. A gait cycle, seen in Figure 2.3, begins when one foot contacts the ground (known as initial contact or heel strike) and ends when the same foot contacts the ground again. Thus, each cycle begins at initial contact with a stance phase in which both feet are on the ground and proceeds through a swing phase, in which only one foot is on the ground, until the cycle ends with the limb's next initial contact. Stance phase accounts for approximately 60 percent, and swing phase for approximately 40 percent, of a single gait cycle. Typical spatiotemporal parameters that are calculated per gait cycle and used for gait assessment are: step length [m], walking velocity [m/s], and cadence [steps/min] as well as the gait cycle time [min] and the percent of the gait in which there is single or double support (Winter, 2005). It is important to note that although human ambulation is considered cyclical, minimal research has validated this assumption.

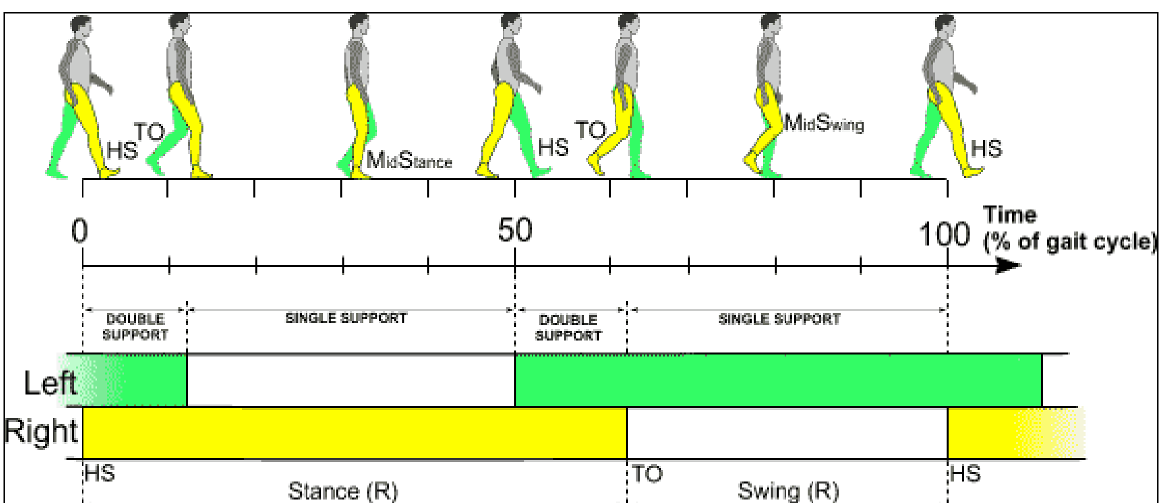

Figure 2.3: Illustration of a typical gait cycle[48].

Kinetics is the general term given to the study of forces, moments and powers that cause movement. The kinetic values are calculated using force plate data. Since force plates typically measure all forces applied to its surface one must ensure only a single foot strikes the force plate, often referred to as a clean strike. For clean strikes, inverse dynamics analysis is used to determine the net joint moments from the resultant GRFs. An example of typical data measured by force plates in the vertical, anterior-posterior (A-P) and medio-lateral (M-L) directions is seen in Figures 2.4-2.6. It is important to note that the M-L force component is known to be very variable, small in magnitude (1/100 of the vertical GRF component), hard to define and therefore rarely used for analysis purposes. As seen in Figure 2.4, in typical gait the vertical force shows a rapid rise at heel contact as full weight bearing takes place. Then, as the knee flexes during midstance, the plate is partially unloaded causing the drop in the vertical force and towards the end of the single foot gait cycle, during push off, the plantar flexion muscles are active causing a second peak in the vertical force. As for the A-P force, at the beginning of the gait cycle the force is negative indicating backward force between the foot and the ground. The force becomes positive near midstance due to the foot pushing back against the plate. The M-L component of the GRFs, although being very variable within a subject as well as among subjects, often shows a short initial reaction force in the medial direction that results from outward moving of the foot during heel strike and landing followed by a reaction force in the lateral direction (White 1998, Nigg 1999). The body’s M-L motion has been shown to be partially stabilized by the M-L foot placement which affects the M-L force, therefore the M-L force may be considered an indicator of stability and balance during walking (MacKinnon, 1993; Simpson, 1999, Donelan, 2004). The ground reaction force vector passes upward from a point called the center of pressure (COP). The GRF causes a torque on the foot segment and subsequent more proximal segments (thigh and pelvis) experience the GRF torque modified by the responses of the more distal joints (Winter, 2005). Theoretically the total GRFs applied at the COP location are also those that effectively act on the center of mass (COM), and should result in similar whole system dynamics that propel the COM forward. It should be noted that the ground reaction forces are the summation of all mass-acceleration products of the body segments and therefore are impacted by the body weight.

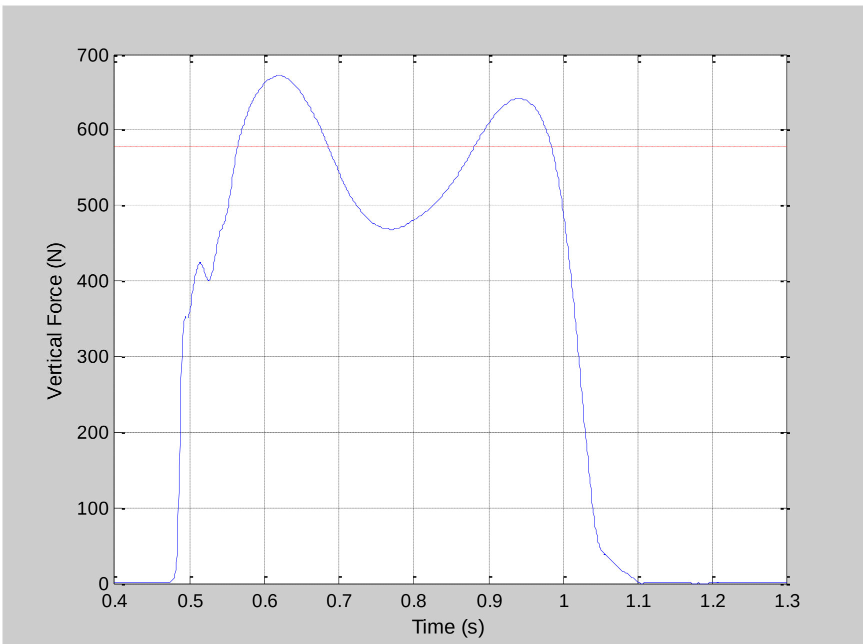

Figure 2.4: Typical pattern of the GRF component in the vertical direction.
The red dashed line represents body weight.

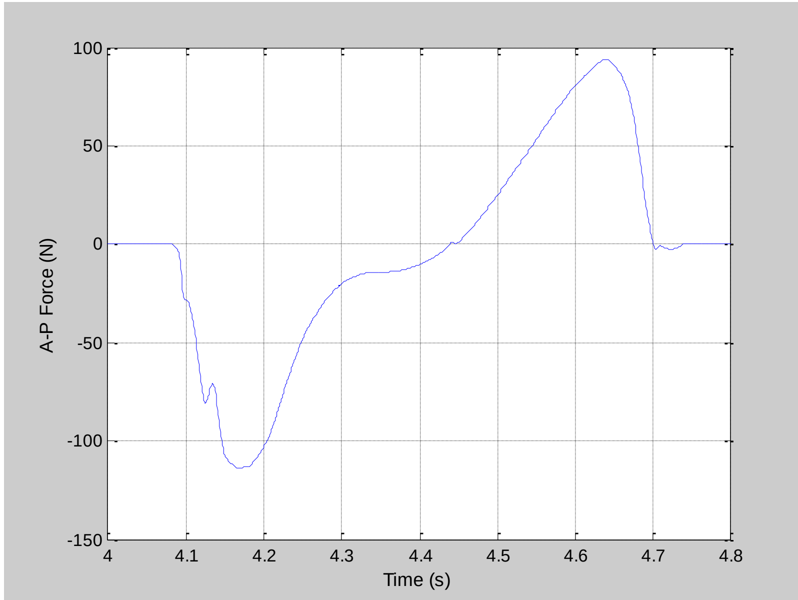

Figure 2.5: Typical pattern of the GRF component in the A-P direction.

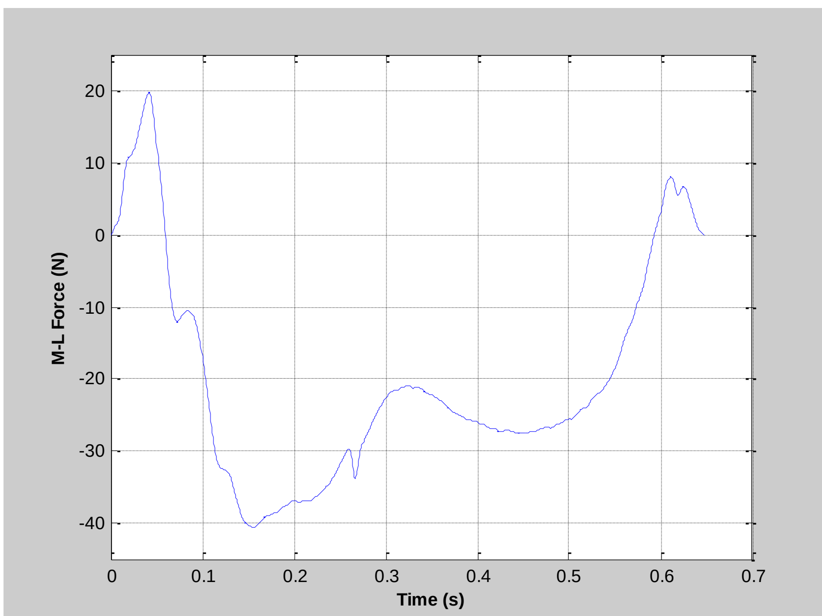

Figure 2.6: Typical pattern of the GRF component in the M-L direction. Positive force value is medial force and negative values are lateral forces.

2.2.1. Existing work in the field of GRFs during gait

Existing work in the field of gait analysis relies on the use of kinematic, anthropometric and kinetic data for inverse dynamics calculation of net joint moments and powers during gait. Yet few existing studies have addressed the stability or consistency of the GRF data during quantitative motion analysis.

Herzog et al. (Herzog, 1989) developed a measure of symmetry/asymmetry for normal human gait which could also be used to quantify symmetry/asymmetry of normal human gait for selected gait variables using force plate data. Asymmetries were quantified using a symmetry index shown in equation (2.1) where X_L and X_R are the gait variables for the left and right legs respectively.

$$SI = \frac{X_R - X_L}{\frac{1}{2}(X_R + X_L)} \cdot 100\% \tag{2.1}$$

Their findings showed human gait is asymmetrical by nature and although the differences between the left and right legs deviated less than ±4%, none of the subjects exhibited perfect (0% deviation) gait symmetry. Herzog et al collected five trials for each foot and averaged the five trials causing a reduction in the variability prior to their analysis of symmetry. It should be noted that although symmetry is rarely found, gait variables are often compared clinically between the left and right sides, and the amount of asymmetry or deviation can be useful.

Kirkpatrick et al. (Kirkpatrick, 1994) studied the reproducibility of the vertical ground reaction force in typically developed children and children with cerebral palsy and found that the vertical ground reaction force has good reproducibility both intrasubject and intraday. They concluded this measurement has a potential role in the objective assessment of medical and surgical interventions.

Steinwender et al. (Steinwender, 2000) examined intrasubject repeatability of gait analysis data, both kinematic and kinetic data, in typically developed children and children with CP. They found that repeatability of kinetic parameters was in general higher than those for the kinematic parameters supporting the belief that GRFs are reliable measurements.

Engsberg et al (Engsberg, 1993) determined normative ground reaction force data for able bodied children and below knee amputee children during walking.
The study, which included 225 typically developed children, showed that for these able bodied children no significant differences existed between the right and left legs, between genders and among different ages.

Seliktar et al. (Seliktar, 1979) developed a gait consistency test based on the time force integral (also referred to as the impulse value) of the A-P force component. The study's approach was based on the assumption that the A-P velocity vector is expected to be equal in two equivalent points of consecutive gait cycles, for instance the velocity of the COM during right heel strike, during two consecutive gait cycles, is expected to be equal and therefore implies that the difference of impulse or momentum between the two points should be zero. If the difference in momentum is zero the difference of forces should be zero as well, as described in equation (2.2).

$$F\Delta t = ma \cdot \Delta t = m\Delta v \rightarrow if \;\; \Delta v = 0 \;\; than F\Delta t = 0 \qquad (2.2)$$

The results of the study, consisting of 28 adult subjects with normal and pathological gait, defined consistency as the sum of the A-P impulse over one full gait cycle divided by the absolute values of impulse both for the braking and propulsion forces. The calculation is described in Figure 2.7 and equation (2.3):

$$Consisteny\,[\%] = \frac{A + B + C + D}{|A| + B + |C| + D} \tag{2.3}$$

As can been seen in the equation (2.3) consistency was checked over two force plate strikes, for both legs, left and right. The use of only one side and the assumption of symmetry is not appropriate for studying pathological gait since in pathological gait many times there are compensations between both legs if one is weaker than the other. When looking at Figure 2.7 and equation (2.3) it is clear that the contribution of impulse 1, the impulse of the right foot from the time of the left foot heel strike to when the right foot toes off, is missing from the equation. This is because most labs have only 2 force plates and in that case the data for impulses 1 and 2 would not be available. The important assumption made in this study was that in typical gait cycles, area 1 and the dashed part of area D should equal each other and in cases where they do not equal each other, the expected error was found to be less than 8%. That is, above a value of 8% the gait cycle was inconsistent.

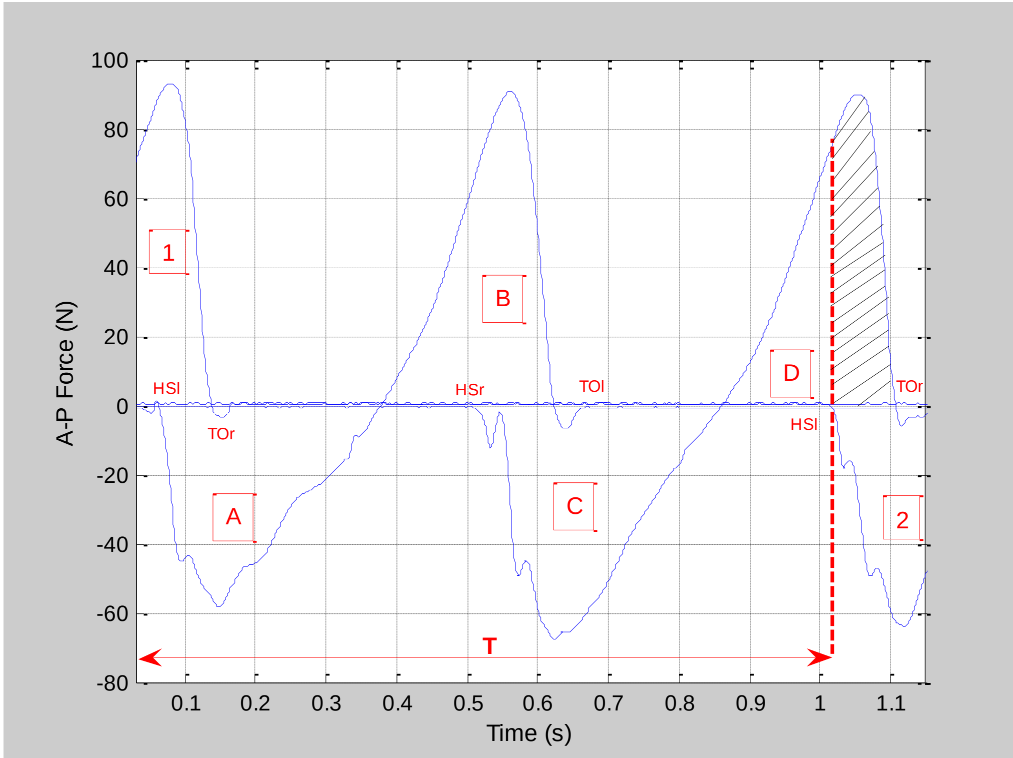

Figure 2.7: The GRFs in the A-P direction during two force plate strikes where T is the gait cycle time, HS and TO are the heel strike and toe off points and ‘l’ and ‘r’ stand for left and right legs.

2.3. Use of accelerometers and foot switches in gait analysis

Acceleration data of body segments can be obtained from direct accelerometry measures, or derived from kinematic data which provides the displacement of chosen body landmarks. Recent advances in the field of accelerometry have demonstrated improved reliability, and coupled with their decreasing cost, direct measures of segment or body acceleration can provide a more useful way to quantify gait characteristics (Mayagoitia, 2002; Zilstra, 2003; Aminian, 2004; Tanaka, 2004; Luinge, 2005).

2.3.1. Foot Switches

One of the temporal divisions in gait analysis in general and GRFs analysis in particular is the gait cycle events. One method of detecting gait events is with the use of Force Sensitive Resistors (FSRs) also known as foot switches. The principle by which foot switches operate is a changing resistance according to the pressure applied. The foot switch is a sensor whose electrical resistance decreases when pressure is applied to the active surface. The rise and fall times of the foot switch's output are on the order of 1 to 2 milliseconds. The foot switch's response approximately follows an inverse power law characteristic such that within the range of the weights of the subjects of this study an approximately linear relationship exists between FSR resistance and the force applied to it (Yaniger, 1991; Smith, 2002). The foot switch, shown in Figure 2.8 is a thin 4 layer elastic material. The 4 layers include:

1. A layer of electrically insulating plastic.
2. A layer of the active area which consists of a pattern of conductors. This layer is connected to the leads on the tail which are charged with an electrical voltage;
3. A layer of plastic spacer which also includes an opening aligned with the active area, as well as an air vent through the tail.
4. A layer of flexible substrate coated with a thick polymer conductive film, aligned with the active area.

When external force is applied to the foot switch the resistive element is deformed against the substrate. Air from the spacer opening is pushed through the air vent in the tail and the conductive material on the substrate comes into contact with parts of the active area. A higher pressure would cause more of the active area to touch the conductive element and to lower the resistance. By placing the footswitches under the toe and heel the gait events of heel strike and toe off can be detected (Yaniger, 1991; Smith, 2002). The pattern of the signal received from a foot switch for a subject walking is illustrated in Figure 2.9.

Figure 2.8: A foot switch.

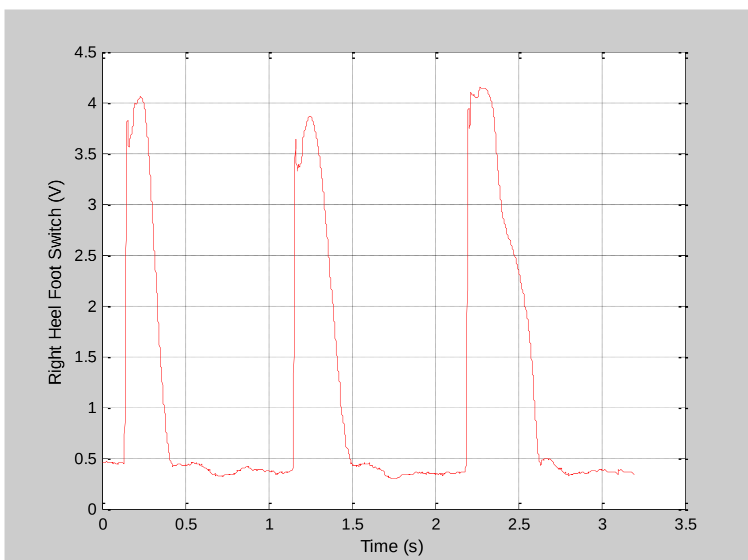

Figure 2.9: A foot switch signal from the heel of a walking subject.

2.3.2. Accelerometers

Accelerometers measure acceleration and gravity induced reaction forces. Single- and multi-axis accelerometers are available and can be used to detect the magnitude and direction of acceleration. Different kinds of accelerometers exist in the market and include strain gauge accelerometers, piezoresistive accelerometers and piezoelectric accelerometers. Strain gauge technology has been reviewed in chapter 2.1, piezoresistive accelerometers are based on the same mechanical principles however, piezoresistive elements in the form of silicon crystals rather than strain gauges are used to convert applied stress to electrical signals (Nigg, 1999; Kavanagh, 2008).

In order to derive meaningful acceleration data using accelerometers, a few steps prior to and post data collection need to be taken to ensure the integrity of the data. Prior to data collection, the accelerometer sensor force-voltage relationship needs to be calibrated for each axis of the accelerometer. Calibration is usually accomplished by aligning each of the accelerometer's axes against gravity to determine its output for gravity ($\pm$g), and then in a non-gravity aligned position to determine the non-gravity dependent bias. Once calibration is done, the accelerometer's sensitivity for each channel can be calculated. Equation (2.4) describes how a channel's sensitivity is calculated:

$$Sensitivity = (V_{+g} - V_{-g}) / 2 \tag{2.4}$$

Where V_{+g}, V_{-g} are the outputs measured against positive and negative gravity orientation respectively. Once the sensitivity and bias values are calculated the output data from the accelerometers can be translated into (m/s^2) units using equation (2.5):

$$
\begin{aligned}
&a\,[m/s^2] = g \cdot (a_V \cdot sensitivity - bias \cdot sensitivity) \\
&g = earth's\ gravity \\
&a\ is\ the\ desired\ acceleration\ in\ m/s^2 \\
&a_V\ \ the\ output\ \ from\ the\ accelerometer.
\end{aligned}
\tag{2.5}
$$

In the case where the accelerometers are tilted off axes during data collection, the tilt must be taken into account in the calculation by measuring the angle of tilt and adding it into the calculations and analysis (Moe-Nilssen, 1998).

2.3.2.1. Existing work in the field of accelerometers use in gait analysis

Accelerometers have been shown to be accurate and consistent in detecting gait parameters in adults and typically developed children. Moe- Nilssen et al. (Moe-Nilssen, 2004) used a triaxial accelerometer, placed on the lower trunk, to measure cadence and step length during timed walking over a range of self-administered speeds in adults. Henriksen et al. (Henriksen, 2004) focused on the test-retest reliability of accelerometers used in gait analysis and concluded they were found to be reliable after comparing 2 sets of data measured during two consecutive days in 20 adult subjects. Brandes et al. (Brandes, 2006) expanded on Moe-Nilssen’s findings by examining whether these results can be received in typically developed children and concluded that spatio-temporal parameters of gait in typically developed children can be estimated using an accelerometer situated at the lower trunk.

In a study performed by Meichtry et al. (Meichtry, 2007) accelerometers placed on the lower trunk were used to compare acceleration measured using an accelerometer to the acceleration calculated from GRFs as well as to assess work and power in able-bodied gait. The results, that looked at the root mean square of the vertical and A-P accelerations showed high correlations between the acceleration calculated from the accelerometer and the acceleration calculated from the GRF data. It should be noted that the accelerometer's location in the study by Meichtry et al. was at the L3 region of the spine while previous studies report the S2 region of the spine as the approximate COM location (Gard, 2004; Bennett, 2005). Despite this reported variation in approximate COM location, both approaches report similar and adequate reproductions of COM mechanics from accelerometry in comparison to calculations based on GRF data.

The use of accelerometers to model COM mechanics provides a basis for an alternative method to assess consistency of gait, GRFs and energy expenditure in typical and pathological populations was provided.

2.4. Gait in Children with Cerebral Palsy

Cerebral Palsy (CP) is caused by faulty development of, or damage to motor areas in the brain that disrupts the brain's ability to control movement and posture. The term “cerebral” refers to the two halves or hemispheres of the brain, in this case to the motor area of the brain's outer layer (the cerebral cortex), the part of the brain that directs muscle movement; “palsy” refers to the loss or impairment of motor function. Although CP affects muscle movement, the primary impairment is not in the muscles or peripheral nerves, rather the cortical centers that control the timing and magnitude of motor force production are impaired. This primary damage to the cerebral cortex, occurring pre- or peri- natally, is static and secondary impairments of the musculoskeletal system occur with some degree of related disability is typically noted (Krigger, 2006; Green, 2007).

2.4.1. Criterions for the different types of CP

CP is typically classified by the body regions affected. Hemiplegic CP refers to the condition when only one side (right or left) of the body is affected, when both the lower extremities are involved the term diplegic CP is used. When all four extremities are affected, the individual is referred to as having quadriplegic CP. Diplegic CP is the most common of all types, occurring in about 45% of children with CP. In addition to classification by affected body region, the type of motor impairment is also used to characterize CP. Spastic CP, in which the muscles often generate a velocity-dependent increase in force production that is not under voluntary control, and increase in muscle tone is noted, is the most common type (Green, 2007).

Children diagnosed with CP are also categorized into levels of functional mobility according to their walking abilities using a Gross Motor Function Classification System (GMFCS). The GMFCS has five levels with level one being the level in which the child walks without any restriction and five being the level in which the child's self mobility is severely limited lacking the basic antigravity postural control (Wood, 2000). It has been shown there is a distinct difference in oxygen cost in the different GMFCS levels and as the severity of involvement increases so does the oxygen cost therefore the higher the level of GMFCS level, the higher the oxygen cost value (Johnston, 2004). Children in GMFCS levels 3, 4 and 5 need an assistive device such as crutches or a walker, and often orthoses to allow or improve their ability to ambulate. A study performed by Maltais et al. (Maltais, 2001) demonstrated that walking with the use of orthotics improve walking efficiency by lowering oxygen cost.

2.4.2. Common movement abnormalities associated with CP

Children with cerebral palsy exhibit a wide variety of symptoms including:

1. Lack of muscle coordination when performing voluntary movements.
2. Stiff or tight muscles and exaggerated reflexes (spasticity).
3. Toe or foot drag during swing phase.
4. Abnormal and often energy inefficient gait patterns.
5. Variations in muscle tone, either too stiff or too floppy.
6. Excessive drooling or difficulties swallowing or speaking.
7. Shaking (tremor) or random involuntary movements.
8. Difficulty with precise motions, such as writing or buttoning a shirt.

The symptoms of CP differ in type and severity from one person to the next, and may even change in an individual over time. Some people with CP also have other medical disorders, including mental retardation, seizures, impaired vision or hearing and abnormal physical sensations or perceptions. CP doesn’t always cause profound disabilities. While one child with severe CP might be unable to walk and need extensive, lifelong care, another with mild cerebral palsy might be only slightly awkward and require no special assistance. Motor level impairments of CP include difficulty with controlling the magnitude and timing of force generation by the involved muscles, difficulty maintaining balance or walking and involuntary movements as well as imbalance between agonist and antagonist muscles across joints also known as co-contraction, all of which result in an increased amount of energy required for ambulation in this population (Crenna, 1998; Rodda 2001).

2.4.3. Common gait patterns in CP

CP results in several common atypical gait patterns according to the severity of involvement, and the muscles affected. The typical pathological gait patterns found in CP include the following (Becher, 2002):

1. A pattern of insufficient foot lift in swing and forefoot landing at initial contact due to insufficient activity in the tibialis anterior and/or shortening of the gastrocnemius muscle is mostly seen in children with mild CP.
2. Children with more involved cases of CP may show insufficient foot lift in mid-swing as well as insufficient knee extension after initial contact, and no heel rise at midstance. This is caused by premature activation of the triceps surae muscle group. The forefoot landing may be caused not only by insufficient foot lift but also by incomplete knee extension in terminal swing due to insufficient selective motor control. This results in a “toe-walking” pattern which often includes a “stiff knee” component.
3. In severely affected children, a gait pattern characterized by hip and knee flexion in midstance can be present (with or without heel rise). This gait pattern can be caused by strong abnormal activity (with or without passive muscle shortening) of

the gastrocnemius and hamstring muscles or by abnormal activity of the psoas and hamstring muscles. These children are at high risk of developing shortening of the psoas, hamstring, and gastrocnemius muscles, and, in later stages, flexion contractures in hip and knee joints. This gait pattern is very energy consuming, and as the child's body mass increases around puberty, deterioration is noted in their ambulation ability and the teens often require wheelchairs for mobility. This pattern is often termed "crouched gait".

A common way to monitor the disorder over time and growth of this population as well as track changes due to interventions is by studying the biomechanics of gait patterns and comparing those patterns and gait parameters to the gait of typical developed children of the same age. Current rehabilitation interventions to improve walking efficiency in children with CP are directed at improving impairments in the neuromusculoskeletal system. Such interventions include muscle strengthening; stretching exercises; serial casting; ankle foot orthoses; pharmacological treatments to improve balance of muscle tone; surgical interventions such as tendon transfers, lengthening of muscles and the use of functional electrical stimulation (FES) systems. Motion analysis and energy expenditure tests are often used to assess the impact of such interventions on gait.

Motion analysis can be used to quantify the kinematics and kinetics of gait (Davis, 2004). Metabolic efficiency is quantified through measurement of oxygen uptake during walking (Bowen, 1998; Stout, 2004). However, quantitative motion analysis with energy expenditure testing take place in a closed lab, outside the everyday environment the children ambulate in, and the complexity of the natural environment may affect the children's walking.

Individuals with the diagnosis of CP are one of the most commonly studied populations in gait analysis. The common pathological gait patterns noted in children with CP and the evidence supporting a relationship between energy expenditure and gait abnormalities provided a basis for the inclusion of a small sample of children with CP in this project to compare the performance of the research methodology between typical gait and pathological gait.

2.5. Metabolic efficiency measurements.

Measuring the metabolic energy required to execute a task is an intuitively appealing way to quantify task efficiency – lower energy requirements imply greater mechanical efficiency. Therapies that reduce the energy demand of a task, including ambulation, are believed to be effective therapies. In the clinical setting, task energy demand is quantified through pulmonary tests that measure oxygen consumption during task execution. Although providing an accepted measure of energy demand (Rose, 1985; Boyd, 1999; Waters, 1999), these tests are technically demanding, staff intensive and the encumbrance of the equipment alone may impact gait in children as can be seen in Figure 2.10. For this reason studies have tried to link energy expenditure (EE) to other aspects of gait, and create models of gait that can predict EE without the use of pulmonary tests.

Figure 2.10: A subject walking during an energy expenditure test.

Previous studies have looked at the relationship between mechanical work as well as other mechanical parameters calculated from GRFs and the metabolic cost of walking. Burdett et al. (Burdett, 1983) showed the existence of high correlation between mechanical work per second walked and the metabolic energy consumption in healthy adults. Other studies have looked at the relationship between the COM's movement and metabolic cost as well as the relationship between mechanical work and metabolic cost (Donelan, 2002; Gordon 2009). No reported studies, that we are aware of, have focused on whether a relationship between metabolic cost and GRFs can be defined, and whether the characteristics of the GRF data and deviations from typical values can explain higher values of metabolic cost.

Chapter 3: Methods

3.1. Parameters of consistency and stability in gait

As previously mentioned, consistency of gait was quantified in adults (Seliktar, 1979) but we have found no similar reported work in children. The first aim of this research was to address this gap, and expand beyond characterizing consistency to include assessment of stability and characterization of a child's gait using ground reaction forces. Changes in ground reaction forces can quantify the changes in a subject's gait due to time, surgeries, orthotics or other interventions.

3.1.1. Consistency of gait in typically developed children and children with CP (Aim 1)

In order to assess consistency in the gait of typically developed children and children with CP, the method used by Seliktar et al. (Seliktar, 1979) was applied to gait data of typically developed children and children with CP. Retrospective data of a sample of 53 ambulatory children who underwent gait analysis testing at Shriners Hospital for Children (SHC) - Philadelphia motion analysis laboratory were used in this analysis of gait consistency. The sample included three groups of children. The first group included typically developed children, 8 males and 8 females, ranging from 7-17 years of age (Mean=11.2 ± 2.1). The second group consisted of children with diplegic CP, 8 males and 8 females, ranging from 9-17 years of age (Mean=12.5 ± 2.0). The third group included children with hemiplegic CP, with either left or right sides affected, 10 males and 11 females, ranging from 10-17 years of age (Mean=12.9 ± 2.3). Two walking trials for each subject were used in the analysis.

Gait analysis was used to capture ground reaction forces during ambulation. The Motion Analysis Laboratory at Shriners Hospital –Philadelphia is equipped with an 8-camera MX-Vicon motion capture system (Vicon Motion systems, Lake Forest, CA) and 4 AMTI force plates. Analog force plate data were collected through the Vicon system. Each subject wore shorts and a T-shirt throughout the evaluation and gait analysis was performed with the subject walking barefoot and walking at his/her freely chosen walking speed along an 8.4-meter level walkway. Force data were collected when the subject traversed the middle 5 meters. Successful walking trials were trials in which the subject placed a single foot on all four force plates. Force plate data in the anterior-posterior plane and vertical plane, collected at 1200Hz, were extracted from the Vicon's c3d files and imported into Matlab (The Mathworks Inc. 6.5, Natick MA) for further analysis. A Matlab program was written to calculate gait consistency, as defined by Seliktar et al. (Seliktar, 1979).

3.1.2. Parameters of stability in gait (Aim 2)

New parameters were calculated from the A-P impulses: landing stability, propulsion effort ratio and the performance imbalance index. The new parameters examined the difference between the two legs during locomotion. In order to be consistent during the comparison, we defined each side as being either the weaker or stronger side by comparing the left and right vertical impulses and finding which is larger. For children with hemiplegic CP the weaker side is clearly the involved side with lower vertical impulses, however, for typically developed children and children with diplegic CP, a weak and strong side may not be always evident despite slight asymmetries and our approach provided some consistency in the definition. Other approaches can be used, such as looking at the left side over the right side, but the important thing is being consistent.

The landing stability ratio was calculated as the ratio between the values of the braking impulses during the gait cycle.

$$\text{Landing Stability Ratio} = \frac{\text{Braking Impulse}_{\text{weaker leg}}}{\text{Braking Impulse}_{\text{stronger leg}}} \quad (3.1)$$

$$(\textit{Optimal values are those closer to } 1)$$

This measure characterizes events during heel strike, as the foot is landing on the ground, and the difference, if any, that exists between the left and right sides. Such a difference may occur if an individual does not feel secure on a certain leg, and lands on that leg with some instability or less force. The landing stability ratio measure quantifies this behavior.

The second measure, propulsion effort ratio, was calculated as the ratio between the propulsion impulses of a gait cycle.

$$\text{Propulsion Effort Ratio} = \frac{\text{Propulsion Impulse}_{\text{weaker leg}}}{\text{Propulsion Impulse}_{\text{stronger leg}}} \quad (3.2)$$

$$(\textit{Optimal values are those closer to } 1)$$

This measure examines the feet as they push off the ground and propel the body forward. Propulsion effort ratio and landing stability both provide more comprehensive characterization of gait, particularly when they are used together.

The performance imbalance index was calculated as the ratio between the absolute impulse values of the left and right legs.

$$\text{Performance Imbalance Index} = \frac{(|\text{Braking Impulse}| + \text{Propulsion Impulse})_{\text{weaker leg}}}{(|\text{Braking Impulse}| + \text{Propulsion Impulse})_{\text{stronger leg}}} \tag{3.3}$$

(Optimal values are those closer to 1)

These values provide information regarding the contribution of each leg to the locomotion of the body and can also be good comparative measures when trying to assess the difference between conditions such as pre and post interventions, as well as with or without orthotics.

Additional characteristics of the vertical and A-P ground reaction forces were examined as well. The peak propulsion and braking force were calculated for all three populations. The vertical ground reaction force has two active peaks as can be seen in Figure 2.4. The two peaks should be close in magnitude and any difference between them may reflect an issue with control of vertical acceleration/deceleration of the body. The active peaks of the braking and propulsion forces were examined for comparison between them as well as for comparison of the values found in gait of individuals with CP to those found in typically developed children. The minimum vertical force that is found between the two peaks mentioned above is related to knee flexion during midstance and can provide important information therefore values of this minimum vertical force were examined as well. In the M-L direction the peak medial and lateral forces were calculated in order to compare among the three groups. The values of all peak forces were normalized to body weight (BW) so comparison among subjects could be made. Normalized impulse values in all three planes were calculated and compared as well. A

Matlab program was written and used to locate the peak forces, calculate the impulse values and normalize the data.

3.2. Deriving ground reaction force characteristics from the acceleration of a calculated COM (Aims 3 and 4)

The ground reaction forces measured by the force plates represent the acceleration and movement of the body's COM. By applying Newton's second law ($F = ma$) to the measured COM's acceleration and multiplying it by the subject's mass the GRF could be calculated. However, the COM's location during ambulation is not a static point since the arrangement of the body segments and mass are re-arranged with movement making the instantaneous COM location difficult to dynamically quantify. It is for that reason that the COM's location has been determined in different ways and has been reported to be approximately at the L3 region of the spine (Meichtry, 2007) or the S2 region of the spine (Gard, 2004; Bennett, 2005). The hypothesis of this research was that the COM's location could be better assessed using a combination of accelerations of several points on the trunk, the body segment with the largest mass, rather than a single marker placed over the presumed location (Winter, 2005).

3.2.1. Subjects who participated in the data collection

Data from 13 typically developed children, 4 males and 9 females, ranging in age from 7-16 (Mean=11.50 ± 2.87) were collected. The sample size was based on selecting a significance level (α) of 0.05 and power (β) of 0.95 which required a sample size of a minimum of 5-6 subjects (Faul, 2007). Allowing for the possibility of considerable variation and to account for data that might not be suitable for analysis, the sample size chosen was doubled to 10 subjects. Data were also collected from 7 children with CP who ranged from 11 to 15 years of age (Mean=12.25 ± 1.89), 3 males and 4 females, who walked without an assistive device. Anthropometric measures for the participating subjects are presented in Table 3.1.

Table 3.1: Anthropometric measures of the subjects who participated in prospective study.

Typically Developed Children				
Subject	Gender	Age	Height (m)	Weight (Kg)
1	M	7	1.27	25.5
2	M	12	1.60	40.1
3	M	10	1.36	30.2
4	M	15	1.74	56.9
5	F	9	1.36	32.4
6	F	11	1.54	38.6
7	F	16	1.64	65.4
8	F	15	1.68	58.97
9	F	15	1.54	44.7
10	F	13	1.61	60.7
11	F	10	1.37	28.8
12	F	12	1.49	34.9
13	F	8	1.20	20.2

Table 3.1: Anthropometric measures of the subjects who participated in prospective study (continued).

Children with CP				
Subject	Gender	Age	Height (m)	Weight (Kg)
1 (Diplegic)	F	11	1.35	38.0
2 (Left Hemiplegic)	M	16	1.73	95.1
3 (Diplegic)	M	12	1.46	30.5
4 (Diplegic)	M	11	1.36	33.5
5 (Diplegic)	F	11	1.47	67.0
6 (Left Hemiplegic)	F	14	1.49	57.7
7 (Diplegic)	F	13	1.53	40.0

3.2.2. Data collection system

Data were collected during gait using a Vicon motion capture system (Vicon Motion systems, Lake Forest, CA), accelerometers (Delsys Inc, Boston, MA), 4 AMTI force plates (Advanced Mechanical Technology Inc., Watertown, MA) and an Actiheart activity monitoring system (Mini Mitter, Sunriver, OR, USA).

The Motion Analysis Laboratory at SHC –Philadelphia, where data collection took place, is equipped with a motion analysis system consisting of an 8-camera MX-Vicon motion capture system and 4 AMTI force plates. The Vicon system was used to collect marker trajectories and the analog signals from the force plates. The markers' data were collected at a frequency of 120Hz, and the force plate data at 1200Hz. Acceleration and foot switches data were collected using a Myomonitor III wireless system (Delsys, Boston MA). The system included the Myomonitor III main unit and two input modules with 8 channels in each module (Figure 3.1). Two 3D accelerometers, two 2D accelerometers and four foot switches were connected to the two input modules (Figure 3.2). A hardware trigger connected to the main unit and was used to start data collection with the Myomonitor system. The system was connected to a data collection computer via a wireless D-Link receiver. The acceleration and foot switch data were collected at 1200 Hz. Once data collection was complete, it was processed using the Myomonitor software, EMGworks Analysis 3.5, and saved as Excel files (Microsoft, Redmond, WA) for further analysis.

Figure 3.1: The Myomonitor III main unit system with the two input modules connected.

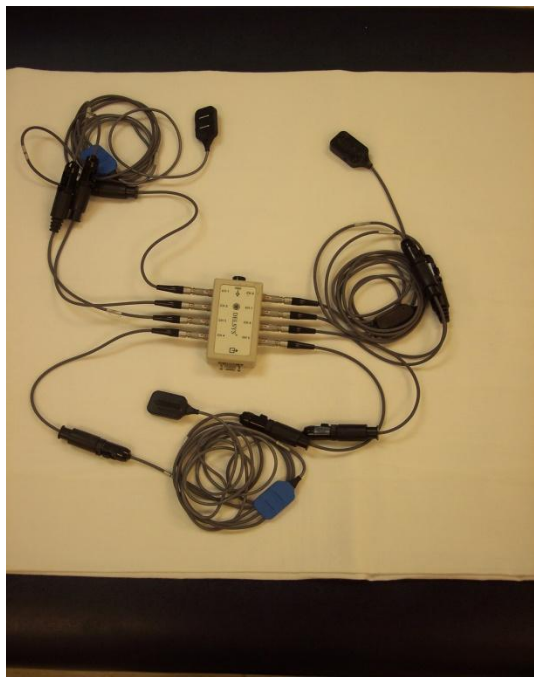

Figure 3.2: An example of an input module with three accelerometers connected to it.

As mentioned previously measurement of metabolic cost or energy expenditure using pulmonary tests indirect calorimetry measures is cumbersome and may affect the subject’s gait. In addition at Shriners motion analysis lab metabolic cost measurements cannot be done while collecting gait data. Therefore, it was decided to use a fairly new system for energy expenditure measurement using the Actiheart device (Figure 3.3). The Actiheart is a device that records physical activity and heart rate with a high level of accuracy by digitizing the ECG signal and calculating the heart rate from the time interval of one peak ECG deflection to the next subsequent peak ECG deflection (true R wave-to-R wave interval). Levels of caloric expenditure can be determined using the information acquired by the device and have been shown to be reliable in adults and typically developed children (Corder, 2005; Crouter 2007). The Actiheart clips onto a single standard ECG electrode with a short ECG lead to another electrode that picks up the ECG signal. It is normally worn on the upper chest. In order to calculate energy expenditure from heart rate using the validated regression equations, the subject’s gender, age, height and weight were recorded.

Figure 3.3: An Actiheart device[49].

3.2.3. Data collection protocol

Testing was performed with the subjects in barefoot condition. For each subject a minimum of two successful walking trials were collected at three walking speeds: 1) the subject's self-selected walking speed; 2) walking faster than their self selected walking speed; and 3) walking slower than typical. Data collection included the following steps:

1. Measurement of the subject's mass, height needed for the Actiheart energy expenditure regression equations.
2. Placement of the Actiheart on the subject's upper left chest.
3. Placement of two 3D accelerometers on their trunk in the sacrum and T3 regions. Two 2D accelerometers were placed on the mid-clavicles to capture shoulders movement during gait. All accelerometers were taped to the subjects back and shoulders (Figures 3.4 and 3.5).
4. Placement of a marker on each accelerometer to allow the movement and orientation of the accelerometers to be verified using the Vicon motion system.
5. Placement of two foot switches on each foot to detect heel strike and toe off. The footswitch data allowed synchronization in time of the force plate and acceleration data.
6. Collection of a static trial in which the subject stood still. The data was used to calculate the accelerometers' tilt.
7. Data collection for the three walking speeds.

Although the Myomonitor system, seen in Figure 3.1, is defined as a wireless system it is a heavy device that could not be carried by the subjects. The wires from the accelerometers and foot switches were also a possible interference with ambulation therefore a laboratory assistant carried the system and ambulated behind the subject. Using the Vicon and Myomonitor software, the resulting kinematic, accelerometer and foot switch data were collected and processed. The results of the static trials were used to calculate the tilt of the accelerometers with respect to the three directions of motion, vertical, A-P and M-L, and used to calculate the true acceleration in those directions.

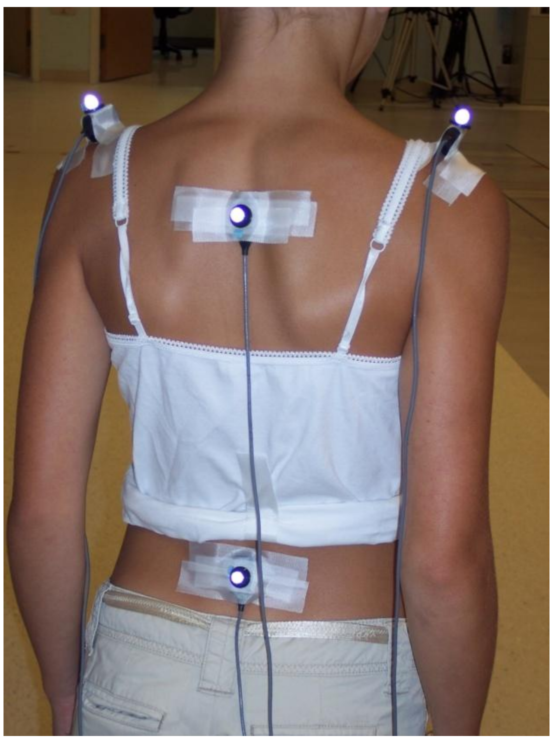

Figure 3.4: A subject with all 4 accelerometers situated on the back.

Figure 3.5: Foot switches as they were applied to the subject’s feet.

3.2.3.1. Analysis of the Actiheart data (Aim 5)

The Actiheart software provides direct values of the energy expenditure values in units of (Kcal/min). The values received for the subjects were compared to the force data and the values derived from it using statistical analysis to show possible relationship between the energy expended during ambulation and the characteristics and measures derived from the GRF data.

3.2.4. Methods for calculating GRFs from acceleration data (Aim 3)

Force plate data provides information for each leg separately which is not the case when looking at the COM. This is the main difficulty when trying to recreate the force curves from the COM's acceleration. Since the COM acceleration does not differ for left and right movements, assumptions had to be made for the double support time. The assumption made was that the force curves could be derived using the acceleration curve patterns during the single support time interval and using the zero force conditions immediately prior to heel strike and after toe off. Equation (3.4) describes the force calculations for the single support phase.

$$\begin{aligned} F_{A-P} &= m \cdot a_{A-P} \\ F_{vertical} &= m \cdot a_{vertical} + m \cdot 9.81 \end{aligned} \tag{3.4}$$

The force curves and data were completed using a cubic spline interpolation function and the known boundary conditions of zero force just before heel strike and after toe off. The cubic spline relied on the known single support data to complete the missing data during double support. The process is illustrated in Figures 3.6 and 3.7.

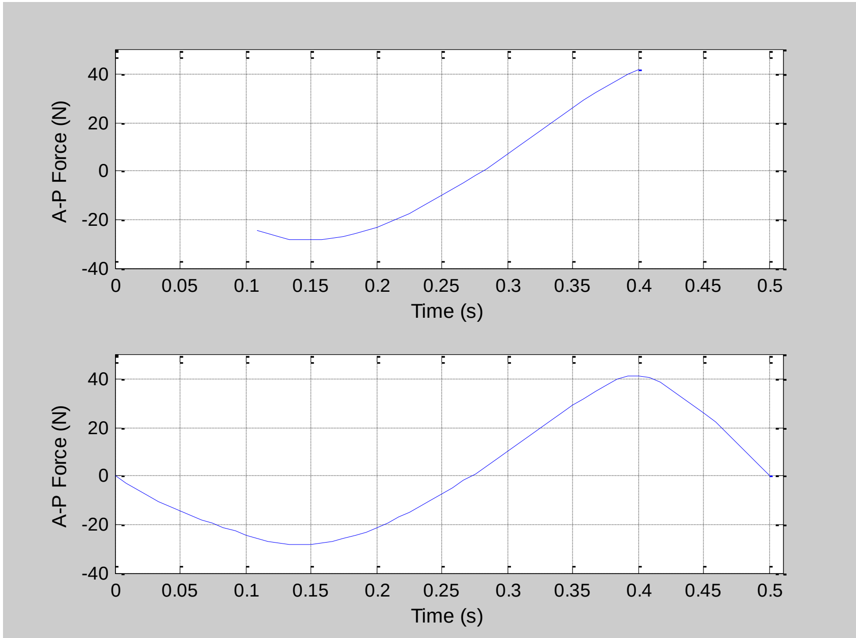

Figure 3.6: Completion of the A-P force curve using a cubic spline function for the double support phases of the gait cycle.

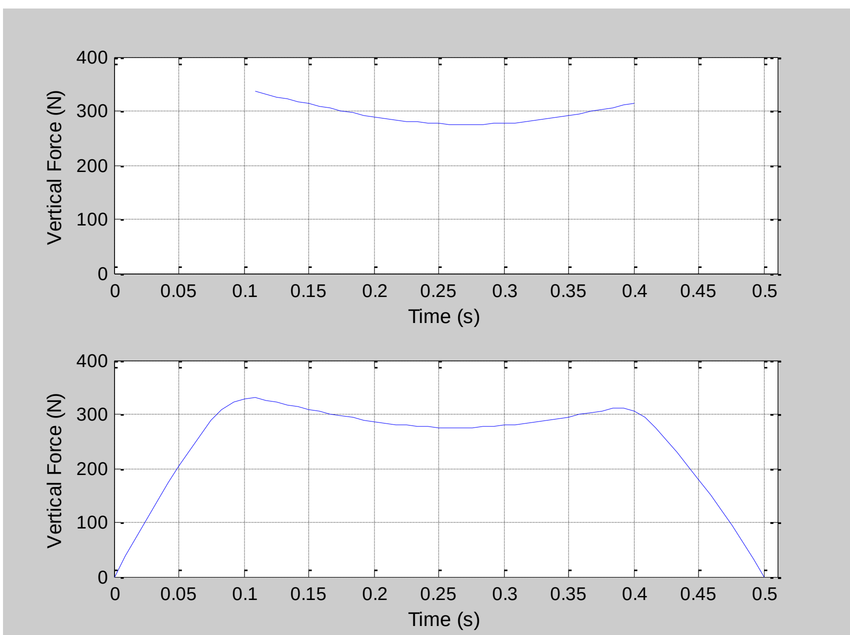

Figure 3.7: Completion of the vertical force curve using a cubic spline function for the double support phases of the gait cycle.

As been previously mentioned, although the acceleration at the sacrum is a good approximation for the COM's acceleration, there are still differences that could be eliminated. For that reason extra accelerometers were placed on the trunk and used with the goal of better predicting the acceleration of the COM.

Although the process mentioned above for the recreation of the GRF data is straight forward and fairly simple, simply taking the acceleration of the sacrum acceleration from the accelerometer and applying the techniques mentioned above would not yield results that provide accurate GRF data. In the attempt of optimizing the results for the prediction of the COM's acceleration two different approaches were taken. The results of the forces received from acceleration values were assessed by comparing certain characteristics of the ground reaction forces from both sets of data. The characteristics chosen for comparison were the maximum propulsion force, maximum braking force, the three characteristics of the vertical GRF: the two peaks and minimum between them and the three impulse values, propulsion, braking and vertical.

3.2.4.1. First approach for the prediction of the COM's acceleration

The first approach for calculating the COM's acceleration focused on data from the four accelerometers placed on the trunk for the creation of a model to predict the acceleration of the COM. This model was based on different combinations of the four accelerometers, providing different contributions to the sacrum acceleration. The hypothesis was that subtle changes to the sacrum's acceleration as modeled by these methods could better predict the COM's acceleration. The models that were tested included:

1. A model that consisted only of the sacrum acceleration.
2. A model that included the sacrum and upper back accelerations.
3. A model consisting of the sacrum and both shoulders accelerations.
4. A model that examined the acceleration of the sacrum and the corresponding shoulder, for example, for the instance where the left leg is stepping on a force plate, the acceleration of the sacrum and left shoulder were taken.
5. A model that examined the acceleration of the sacrum and the opposite shoulder, for example, for the instance where the left leg is stepping on a force plate, the acceleration of the sacrum and right shoulder were taken.

For each of the models, the average acceleration was derived from the data of the accelerometers identified for that grouping.

3.2.4.2. Second approach for the prediction of the COM's acceleration

The second approach relied on the findings of Lee et al (Lee, 2007). In their work Lee et al. showed they could assess the COM's acceleration by placing an accelerometer on the sacrum and using a 60th order finite impulse response (FIR) low pass filter to analyze the acceleration signal. The focus, in this approach, was on the method of filtering of the sacrum's acceleration. The common filtering method used in the analysis of accelerometer data is a Butterworth filter (Winter, 2005) which is an infinite impulse response filter and therefore, using a high order FIR filter that causes significant smoothing of the signal is a novel approach. Lee et al (Lee, 2007) study was performed on adults and used a cutoff frequency of 5Hz. When applied to the data in this work, a cutoff frequency of 5Hz did not produce the significant amount of smoothing shown by Lee et al. therefore it was decided to examine this method with cutoff frequencies of 4Hz and 3Hz. The differences between the Butterworth and FIR filters at cutoff frequencies of 5Hz, 4Hz and 3Hz can be seen in Figures 3.8-3.10.

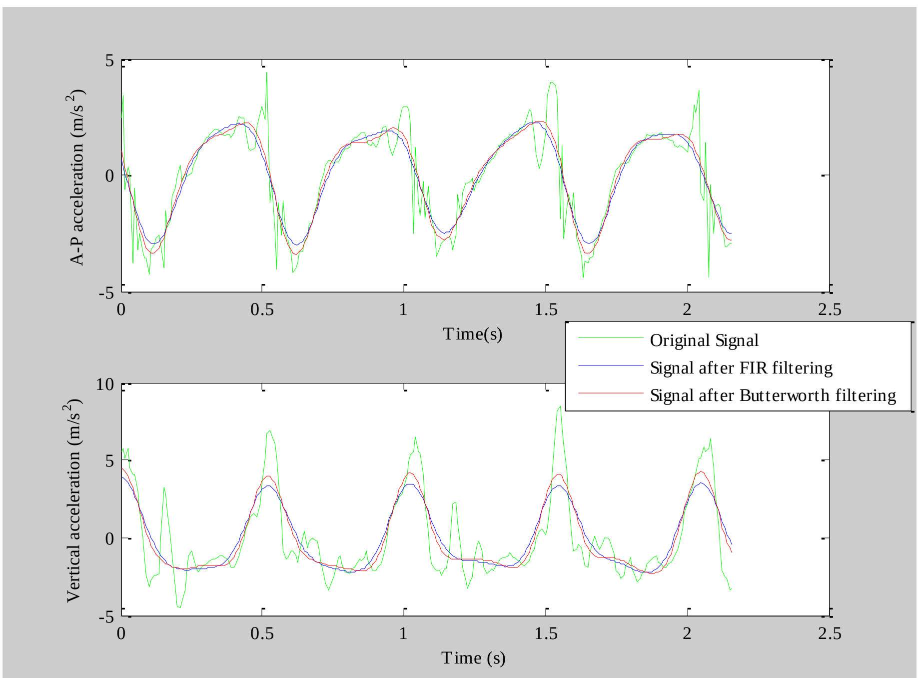

Figure 3.8: Original accelerometer signal compared to the signal after a 4th order Butterworth and 60th order FIR filters at a cutoff frequency of 5Hz

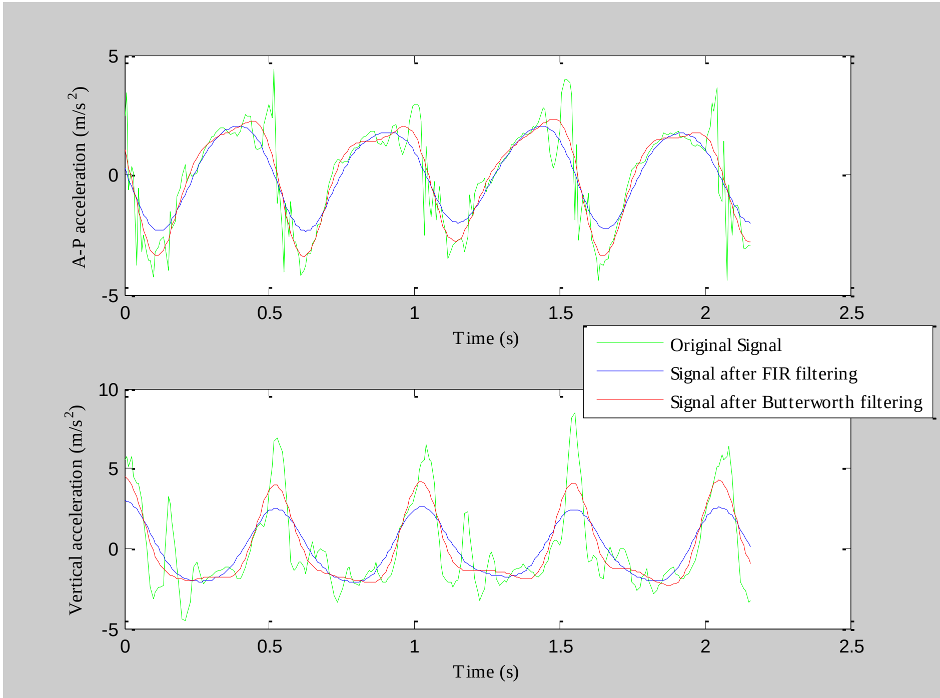

Figure 3.9: Original accelerometer signal compared to the signal after a 4th order Butterworth and 60th order FIR filters at a cutoff frequency of 4Hz.

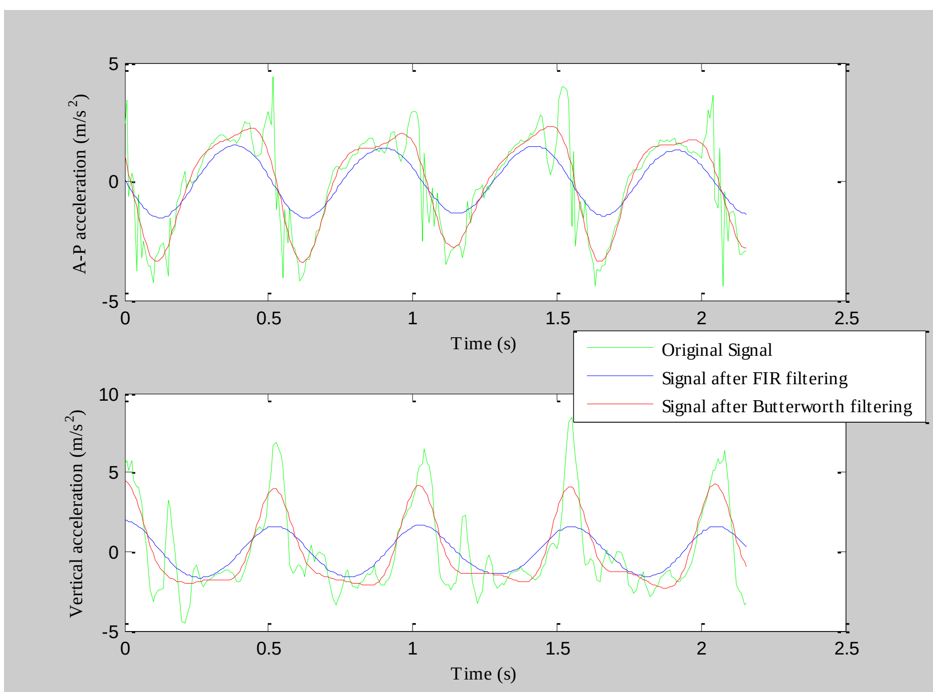

Figure 3.10: Original accelerometer signal compared to the signal after a 4th order Butterworth and 60th order FIR filters at a cutoff frequency of 3Hz.

3.3. Statistical analysis

Statistical analysis was performed on the parameters' values derived from both the GRFs data and from the accelerometers data.

3.3.1. Statistical analysis of the parameters derived directly from GRF data

Descriptive statistics of central tendency were calculated for all dependent variables for each subject group. The seventeen parameters derived from GRF data were divided into two sets. Fourteen of the parameters included in the analysis were parameters that were calculated for each leg and therefore were put in one set. The other three were parameters that looked at the relationship between the two legsand included the landing stability ratio, propulsion effort ratio and performance imbalance index. Both sets of parameters underwent the same analysis done with the exception that the first parameter set was investigated for 4 different populations: 1) typically developed children, 2) children with diplegic CP, 3) children with hemiplegic CP -affected side, and 4) children with hemiplegic CP - less- affected side.

For the second set of parameters, only three populations were present since the hemiplegic population was not divided into more or less affected sides for these parameters. .

Mean values for each parameter for the different populations were determined and assessed for statistical differences between groups. A one-way analysis of variance (ANOVA) that determined any significant differences between variables and a Benferroni post hoc analysis were performed. The analysis was carried out separately for each force parameter where the factor used in the analysis was the population and the levels within the factor being the 3 or 4 different clinical populations.

The second analysis was an analysis of correlation amongst dependent variables using Pearson moment product correlation test with a significance criterion of $\alpha \leq 0.05$. Correlation was performed within each population to examine the relationship between the different parameters, check for redundant information within these parameters (Portney, 1993).

3.3.2. Statistical analysis of the GRFs calculated from accelerometer data

Statistical analysis of the force data derived from acceleration data included correlation analysis using the Pearson moment product correlation with a significance criterion of $\alpha \leq 0.05$ between the parameters derived from acceleration data and the values calculated directly from the force plate data. Prior to the correlation test an ANOVA was run to examine the effects of the different factors in the data since data were collected on four different plates during the subjects walks. The expectations were for a high correlation between the corresponding values calculated from GRF data and accelerometer data.

Chapter 4: Results

The first aim of this research was to investigate and define consistency of gait in typically developed children and children with CP. All other analyses would be based on consistent trials only.

4.1. Consistency test results (Aim 1)

Figure 4.1 presents the results for the consistency test. Consistency was calculated for the three children groups: 1) typically developed children (n=16), 2) sixteen children with diplegic CP (n=16) and 3) children with hemiplegic CP (n = 21), as well as calculated for the entire group of all subjects by applying equation 2.3 to the data of all three groups. Consistency should be present in all populations, pathological or not, and as seen in Figure 4.1, the average consistency values of all three populations are within a 1% range. The average consistency value for the entire children population was close to the findings of Seliktar et al. (Seliktar, 1979) and shows consistency is achieved when the consistency value calculated for a subject is below 6.5%. Quantitative results for Figure 4.1 can be found in Table A1, Appendix A.

The consistent value determined in the test was applied to further analyses and therefore results and conclusions were derived from consistent gait cycles only.

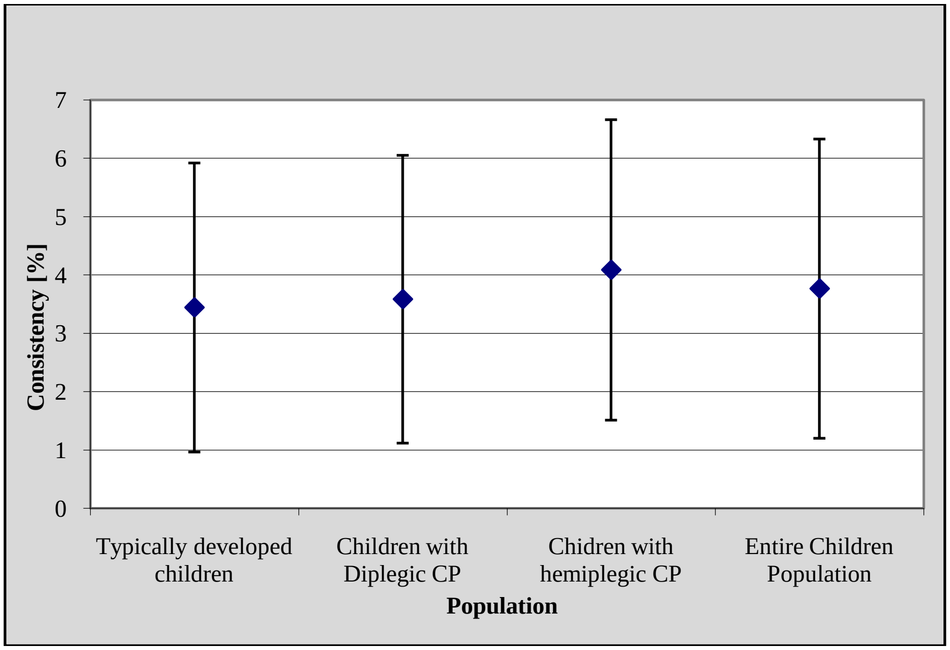

Figure 4.1: Results of the consistency test for typically developed children and children with diplegic and hemiplegic CP.

4.2. Use of ground reaction force data for characterization of gait (Aim 2)

Consistent gait data for the three populations of typically developed children and children with diplegic and hemiplegic CP were used to examine and define the differences between the characteristics of GRFs and parameters derived from them. A summary of the parameters analyzed is presented in Table 4.1.

Table 4.1: Summary of the GRF parameters examined in this work.

Parameter	***Explanation***
Maximum Propulsion Force	The largest force during the anterior directed ground reaction force
Maximum Braking Force	The largest force during the posterior directed ground reaction force
First Maximum Vertical Force	The first peak force in the vertical direction
Second Maximum Vertical Force	The second peak force in the vertical direction
Minimum Vertical Force	The minimum vertical force between the two vertical peak forces
Maximum Lateral Force	The largest force during the lateral directed ground reaction force
Maximum Medial Force	The largest force during the medial directed ground reaction force
Braking Impulse	The integration of the force and time during the posterior directed ground reaction force
Propulsion Impulse	The integration of the force and time during the anterior directed ground reaction force
Vertical Impulse	The integration of the force and time during the vertical directed ground reaction force
Lateral Impulse	The integration of the force and time during the lateral directed ground reaction force
Medial Impulse	The integration of the force and time during the medial directed ground reaction force
Propulsion over Braking Impulse	Ratio of propulsion impulse to braking impulse
Braking over Propulsion time	The ratio of time spent in the braking phase to time spent in the propulsion phase during one force plate strike
Landing Stability Ratio	The ratio of braking impulses of the two legs during a gait cycle
Propulsion Effort Ratio	The ratio of propulsion impulses of the two legs during a gait cycle
Performance Imbalance Index	The ratio of the sum of absolute values of the braking and propulsion impulses between the two legs during a gait cycle.

4.2.1. Examination of the difference in force characteristics between the different populations

The ground reaction force curves in the vertical, M-L and A-P direction were examined in typically developed children (Engsberg, 1993) to define normative values but have not been compared to the values in children with CP. Tables 4.2- 4.4 present the results for typically developed children and children with diplegic and hemiplegic CP. The values examined were the maximum brake and propulsion forces and the duration of those phases as well as the two peak forces in the vertical directions, the minimum vertical force between them and the maximum lateral and medial forces. The force values were normalized to BW to allow comparison among subjects and temporal values were normalized to the gait cycle time (T in Figure 2.6). As would be seen throughout the analysis, in hemiplegic population the results were divided into two groups, the affected side and the non affected side since in this case there is significant difference between the two sides.

It should be noted that although the peak lateral and medial forces are close in value (Table 4.4), especially in typically developed children, for most of the gait cycle the foot applies a medial force (the corresponding ground reaction force is lateral) as seen in Figure 2.6 and quantified by the impulse values presented in the next section.

Table 4.2: Results of the peak and minimum force values in the A-P and vertical directions.

	Maximum braking force [%BW]	***Maximum propulsion force [%BW]***	***First Maximum of the vertical force [%BW]***	***Second Maximum of the vertical force [%BW]***	***Minimum vertical force between the two peaks [%BW]***
Typically Developed Children	-18.99 ± 4.77	22.08 ± 3.50	114.74 ± 10.81	110.92 ± 7.52	76.66 ± 8.11
Children with Diplegic CP	-25.64 ± 9.14	18.54 ± 5.81	133.46 ± 21.16	104.71 ± 16.39	72.20 ± 13.91
Children with Hemiplegic CP-non affected side	-19.24 ± 6.02	21.18 ± 4.32	118.43 ± 17.43	112.82 ± 10.98	71.28 ± 11.55
Children with Hemiplegic CP-affected side	-20.07 ±7.05	15.40 ± 4.48	114.05 ± 15.48	100.65 ± 8.77	75.02 ± 11.01

Table 4.3: Results of the force characteristics in the A-P direction.

	Brake phase time [% gait cycle time]	***Propulsion phase time [% gait cycle time]***	***Brake phase time/Propulsion phase time***
Typically Developed Children	33.23 ± 3.77	28.65 ± 3.51	1.19 ± 0.28
Children with Diplegic CP	27.16 ± 6.14	36.84 ± 6.59	0.79 ± 0.29
Children with Hemiplegic CP-non affected side	31.74 ± 6.81	32.38 ± 6.38	1.03 ± 0.33
Children with Hemiplegic CP- affected side	27.84 ± 6.20	32.53 ± 6.41	0.89 ± 0.29

Table 4.4: Results of the force characteristics in the M-L direction.

	Maximum medial force [% BW]	***Maximum lateral force [% BW]***
Typically Developed Children	5.17 ± 2.26	5.82 ± 1.73
Children with Diplegic CP	6.12 ± 4.86	9.24 ± 3.84
Children with Hemiplegic CP-non affected side	4.97 ± 2.56	7.07 ± 2.78
Children with Hemiplegic CP- affected side	6.17 ± 3.98	7.21 ± 2.92

4.2.2. Examination of the difference in impulse values between the different populations

The ratio of propulsion impulse to braking impulse was examined. The results are presented in Figure 4.2 and Table A2 in appendix A. Values should be close to 1, suggesting symmetry or equal contribution to braking and propulsion actions of the leg during ambulation. The group of children with hemiplegic CP – affected leg exhibited the closest value to 1 but with high variability. The non affected side in children with hemiplegic CP had the value farthest from 1 indicating the most difference between the brake and propulsion phases.

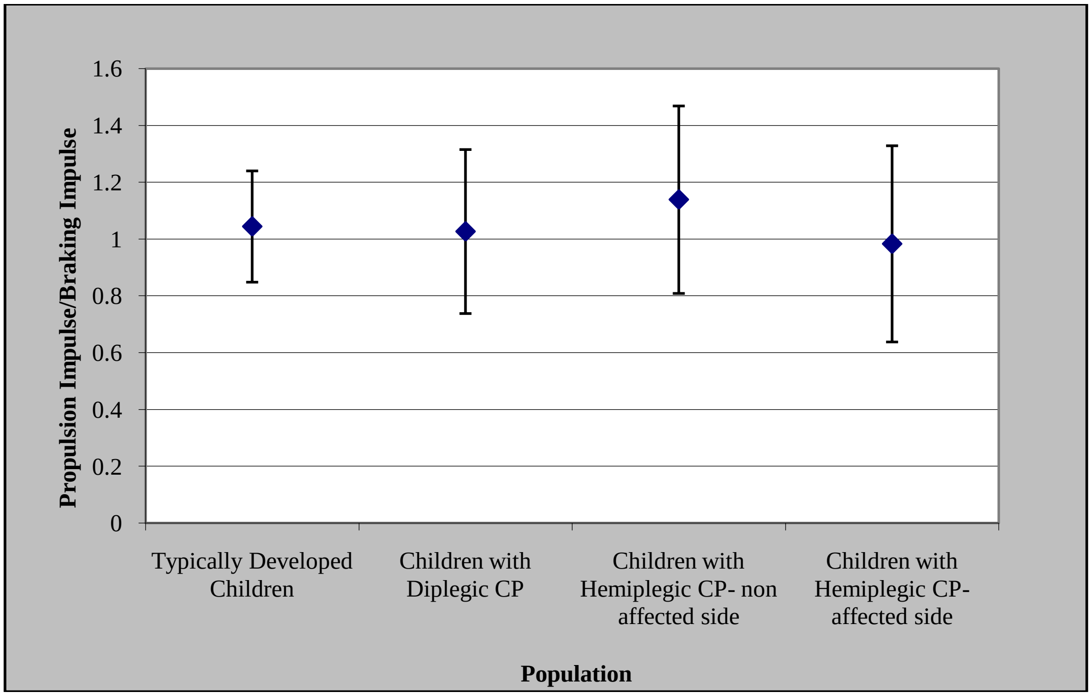

Figure 4.2: Results for the ratio of propulsion to braking impulse. Results are presented as the mean with the standard deviation.

The normalized impulses of the vertical and M-L components of the GRF were calculated as well. The impulses' values, normalized to the subject's weight and gait cycle time (T in Figure 2.6) were calculated for all populations and are presented in Table 4.5. The vertical impulse values for typically developed children as well as children with diplegic CP and the non affected side of children with hemipleigic CP are very similar. When comparing the non-affected to the affected side in children with hemiplegic CP there is a clear difference, with the non affected side generating higher impulses.

Table 4.5: Comparison of the normalized vertical and M-L impulses.

	Vertical normalized impulse	***Medial normalized impulse***	***Lateral normalized impulse***
Typically Developed Children	0.50 ± 0.04	0.0027 ± 0.0015	0.014 ± 0.007
Children with Diplegic CP	0.52 ± 0.06	0.0024 ± 0.0031	0.024 ± 0.012
Children with Hemiplegic CP-non affected side	0.53 ± 0.03	0.0026 ± 0.0022	0.018 ± 0.010
Children with Hemiplegic CP-affected side	0.47 ± 0.02	0.0022 ± 0.0015	0.017 ± 0.010

In another analysis, comparison of the braking and propulsion impulses generated from forces and gait time normalized (to BW and gait cycle time, T) to un-normalized data for all subjects was carried out. The results are presented in Figures 4.3-4.6 and Tables A3 and A4 in Appendix A. The comparison shows the normalization process reduces the amount of variability within a population however; the underlying data trend does not change with the impulse values for both for braking and propulsion being similar for typically developed children, children with diplegic CP and the non affected side in children with hemiplegic CP. In children with hemiplegic CP, the difference between the legs is visible and the involved leg has smaller values both for braking and propulsion impulses whether the values are normalized or not. As in previous results, the variability among children with CP is larger than that found in typically developed children.

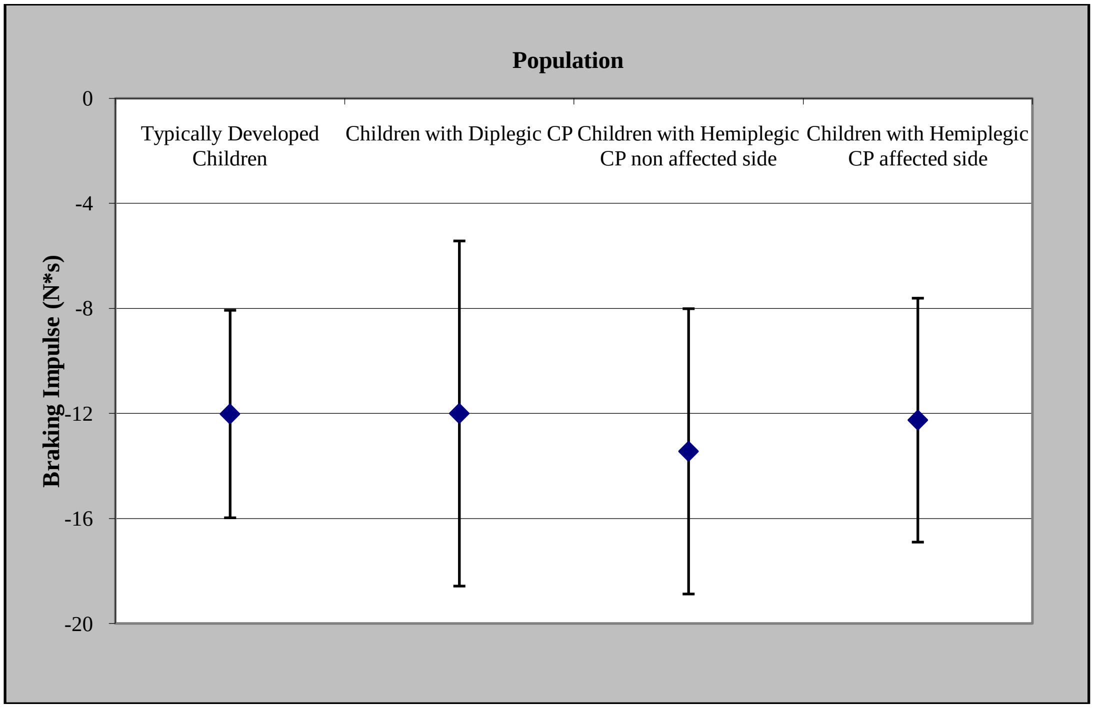

Figure 4.3: Comparison of the braking impulse in typically developed children and children with diplegic and hemiplegic CP. Results are presented as the mean with the standard deviation.

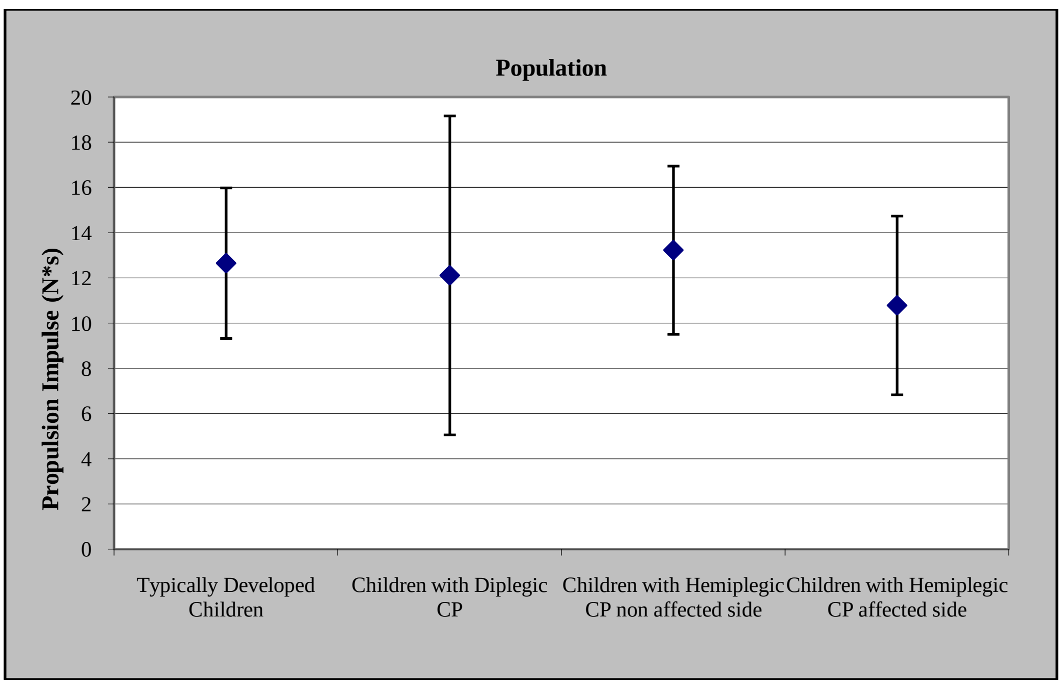

Figure 4.4: Comparison of the propulsion impulse values in typically developed children and children with diplegic and hemiplegic CP. Results are presented as the mean with the standard deviation.

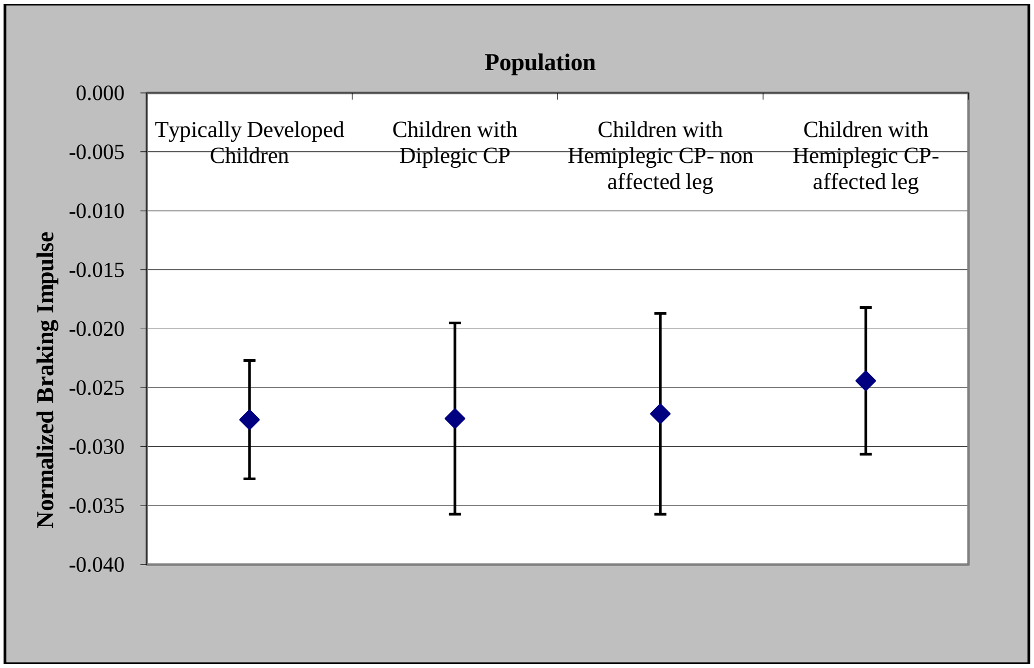

Figure 4.5: Comparison of the normalized braking impulse values in typically developed children and children with diplegic and hemiplegic CP. Results are presented as the mean with the standard deviation.

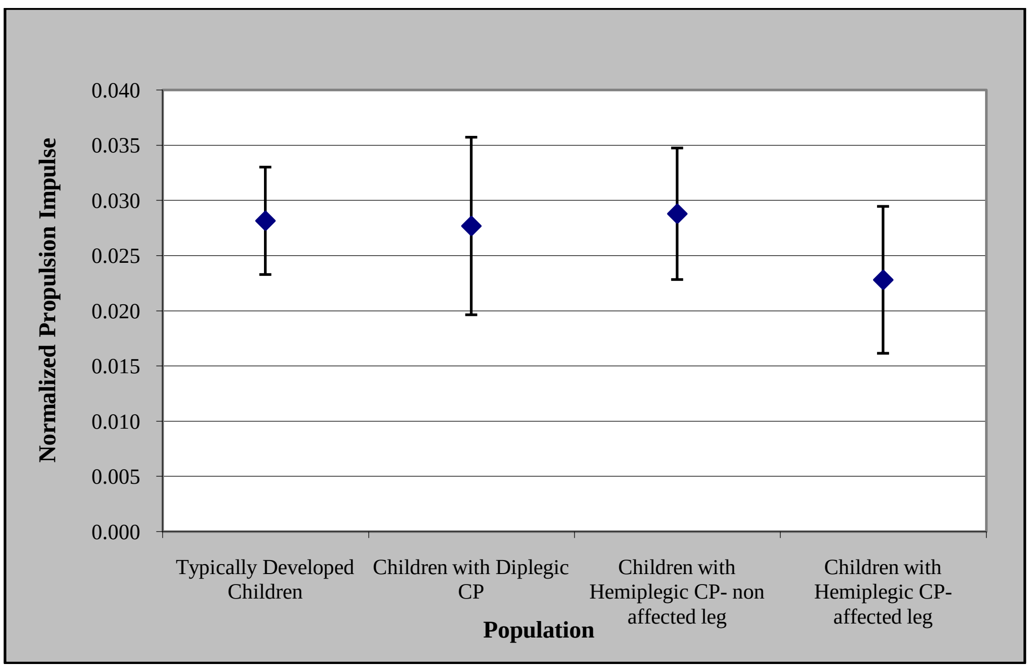

Figure 4.6: Comparison of the normalized propulsion impulse values in typically developed children and children with diplegic and hemiplegic CP. Results are presented as the mean with the standard deviation .

4.2.3. Results for the parameters of stability derived from GRFs

Figures 4.7-4.9 present the results for the calculations of the propulsion effort ratio, the landing stability ratio and the performance imbalance index. Tables A5 and A6 in Appendix A present the quantitative results for Figures 4.7-4.9. As would be expected, overall, typically developed children have less variability and demonstrate similar values from the two legs. Children with CP show more variability having large variability both in the landing and propulsion phases. The performance imbalance index results show a trend with values for children with diplegic CP being lower than typically developed children but not as low as children with hemiplegic CP. These measures provide an opportunity to examine variability and repeatability over a gait cycle with both legs examining the relationship between the two.

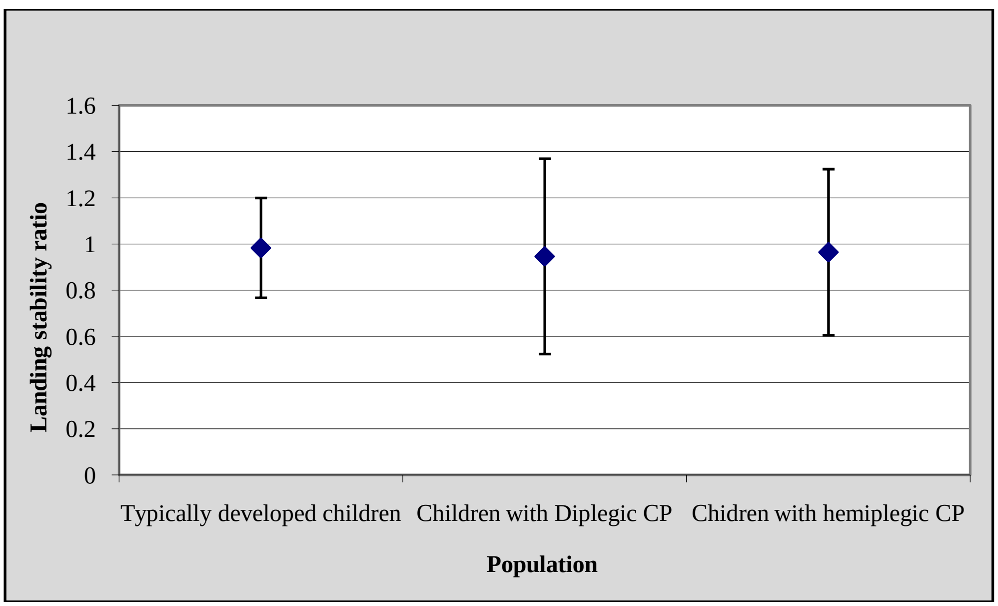

Figure 4.7: Results of the landing stability ratio calculations for typically developed children and children with diplegic and hemiplegic CP. Results are presented as the mean with the standard deviation.

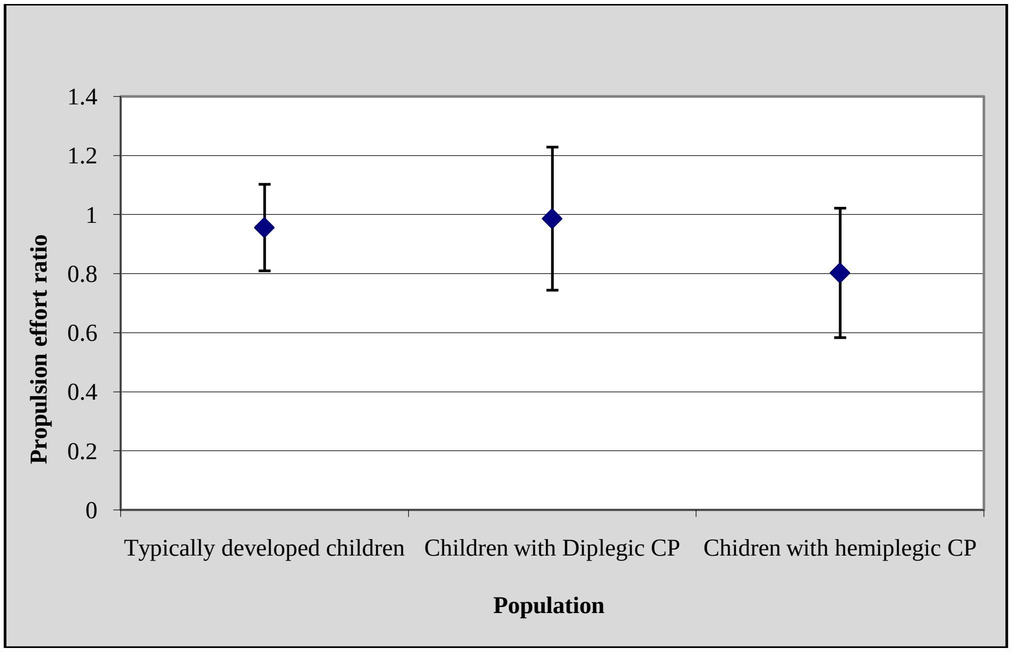

Figure 4.8: Results of the propulsion effort ratio calculations for typically developed children and children with diplegic and hemiplegic CP. Results are presented as the mean with the standard deviation.

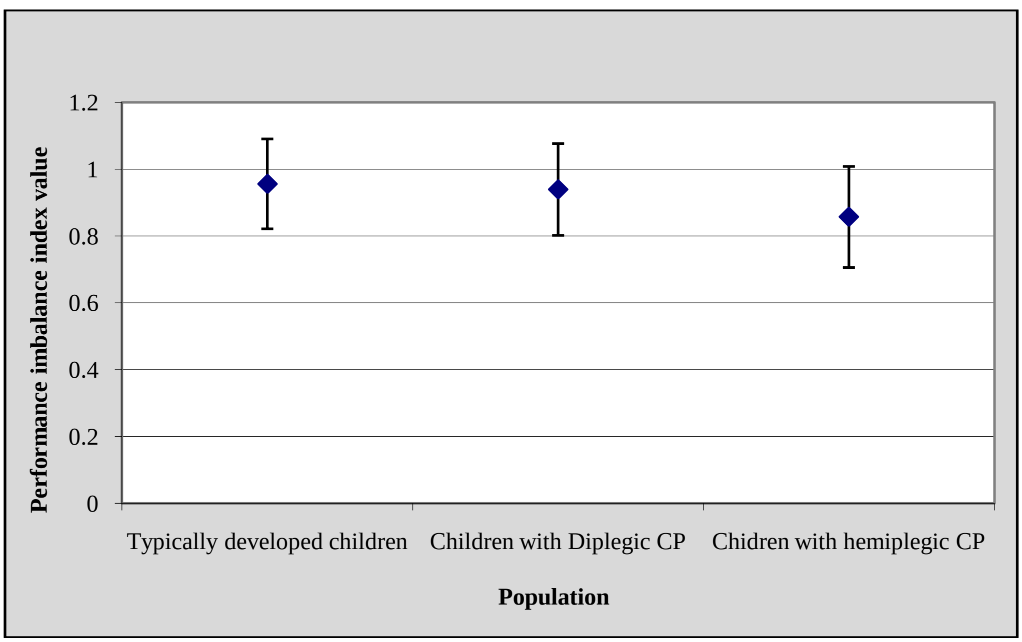

Figure 4.9: Results of the performance imbalance index calculations for typically developed children and children with diplegic and hemiplegic CP. Results are presented as the mean with a the standard deviations.

4.2.4. Statistical results for the force derived parameters

Fourteen different force derived parameters were calculated for the four different children groups: typically developed, children with diplegic CP and children with hemiplegic CP both for the affected side and the non affected side.

The results for the one-way ANOVA are presented in Table 4.6.

Table 4.6: Results for the one-way ANOVA analysis of the fourteen parameters derived from the GRFs data.

	Children with Diplegic CP	***Children with Hemiplegic CP-non affected side***	***Children with Hemiplegic CP-affected side***
Maximum Propulsion Force	TD, H	Di, Ha	TD, H
Maximum Braking Force	TD, H, Ha	Di	Di
First Maximum Vertical Force	TD, H, Ha	Di	Di
Second Maximum Vertical Force	TD, H	Di, Ha	TD, H
Minimum Vertical Force	--	--	--
Maximum Lateral Force	TD, H, Ha	Di	Di
Maximum Medial Force	--	--	--
Braking Impulse	--	--	TD
Propulsion Impulse	Ha	Ha	TD, Di, H
Vertical Impulse	Ha	TD, Ha	TD, Di, H
Lateral Impulse	TD, H, Ha	Di	Di

Table 4.6: Results for the one-way ANOVA analysis of the fourteen parameters derived from the GRFs data (continued).

	Children with Diplegic CP	***Children with Hemiplegic CP-non affected side***	***Children with Hemiplegic CP-affected side***
Medial Impulse	--	--	--
Propulsion over Braking Impulse	--	--	--
Braking over Propulsion time	TD, H	Di	TD

Table 4.6 Notes:
1. "TD" represents significant difference from typically developed children values.
2. "Di" represents significant difference from children with diplegic CP.
3. "H" represents significant difference from the non affected side of children with hemiplegic CP
4. "Ha" represents significant difference from the affected side of children with hemiplegic CP.

Significant correlation coefficients values between the 14 parameters for the four populations are presented in Table 4.7.

Table 4.7: Significant results (p≤0.05) for the Pearson moment product correlation test.

	Typically developed children	*Children with Diplegic CP*	*Children with Hemiplegic CP-non affected side*	*Children with Hemiplegic CP-affected side*
Maximum Propulsion Force and Maximum Braking Force	-0.65	-0.67	-0.59	-0.39
Maximum Propulsion Force and First Maximum Vertical Force	0.52	--	0.57	--
Maximum Propulsion Force and Second Maximum Vertical Force	0.50	0.60	0.47	--
Maximum Propulsion Force and Minimum Vertical Force	-0.54	0.54	0.63	-0.39
Maximum Propulsion Force and Braking Impulse	-0.73	-0.53	-0.60	-0.50
Maximum Propulsion Force and Propulsion Impulse	0.91	0.54	0.84	0.90
Maximum Braking Force and Braking Impulse	0.81	0.43	0.85	0.67
Maximum Braking Force and Propulsion Impulse	-0.73	-0.42	-0.62	-0.46
First Maximum Vertical Force and Maximum Braking Force	-0.76	--	-0.83	-0.53
First Maximum Vertical Force and Braking Impulse	-0.70	--	-0.64	--
First Maximum Vertical Force and Propulsion Impulse	0.57	--	0.54	--

Table 4.7: Significant results (p≤0.05) for the Pearson moment product correlation test (continued).

	Typically developed children	***Children with Diplegic CP***	***Children with Hemiplegic CP-non affected side***	***Children with Hemiplegic CP-affected side***
Minimum Vertical Force and Maximum Braking Force	0.68	-0.32	0.77	0.51
Minimum Vertical Force and First Maximum Vertical Force	-0.61	--	-0.80	-0.81
Minimum Vertical Force and Second Maximum Vertical Force	-0.64	0.71	-0.53	--
Minimum Vertical Force and Braking Impulse	0.60	-0.43	0.71	0.33
Maximum Lateral Force and Lateral Impulse	0.81	--	0.77	0.77
Maximum Medial Force and Medial Impulse	0.58	--	0.62	0.58
Braking Impulse and Propulsion Impulse	-0.84	-0.95	-0.55	-0.46

4.2.5. Statistical results for the stability parameters

In the case of the stability parameters which compare sides, analysis involved only 3 groups as previously stated: 1) typically developed children, 2) children with diplegic CP and 3) children with hemiplegic CP. The one-way ANOVA analysis results are presented in Table 4.8.

Table 4.8: Results for the one-way ANOVA analysis for the heel strike, ambulation effort ratio and performance imbalance index parameters.

	Children with Diplegic CP	***Children with Hemiplegic CP***
Landing Stability Ratio	--	--
Propulsion Effort Ratio	--	TD, Di
Performance Imbalance Index	--	TD, Di

Table 4.8 Notes:
1. “TD” represents significant difference from typically developed children values.
2. “Di” represents significant difference from children with diplegic CP.

The second analysis examined the correlation between the three stability parameters within each population. The results, presented in Table 4.9, support the use of all three parameters and show although there are moderate correlations between the parameters, no correlations are so high that a significant portion of the variability in each parameter is not accounted and each parameter represents different information.

Table 4.9: Results for correlation analysis for the heel strike, ambulation effort ratio and performance imbalance index parameters within the three populations.

Typically Developed Children			
	Performance Imbalance Index	***Landing Stability Ratio***	***Propulsion Effort Ratio***
Performance imbalance index	--	0.46	--
Landing Stability Ratio	0.46	--	-0.54
Propulsion Effort Ratio	--	-0.54	--
Children with Diplegic CP			
	Performance Imbalance Index	***Landing Stability Ratio***	***Propulsion Effort Ratio***
Performance imbalance index	--	0.54	0.57
Landing Stability Ratio	0.54	--	--
Propulsion Effort Ratio	0.57	--	--
Children with hemiplegic CP			
	Performance Imbalance Index	***Landing Stability Ratio***	***Propulsion Effort Ratio***
Performance imbalance index	--	0.66	--
Landing Stability Ratio	0.66	--	-0.49
Propulsion Effort Ratio	--	-0.49	--

4.3. Illustration of the effectiveness of the techniques within subjects

These techniques of GRF derivation and analyses were applied to data from six children with Cerebral Palsy. (three case studies are presented in sections 4.3.1-4.3.3 and the remaining three can be found in Appendix B). Clinical gait analysis requires comparison of subjects over time, pre and post surgical intervention and with and without orthotics. The application of the findings of section 4.2 to six cases demonstrate the GRF measures sensitivity to gait deviations, and provide evidence for the utility and validity for their use of in clinical gait analysis.

4.3.1. Case study one

The first case study was of a female subject with diplegic CP who was seen for a gait analysis 11 months pre (age 11) and 11 months post (age 13) a bilateral hamstring lengthening and application of long leg cast. The comparison is presented in two ways, representative ground reaction forces, normalized to body weight, are presented in Figures 4.10 – 4.12 and the quantitative values and calculated parameters pre and post intervention can be seen in Table 4.10.

Prior to the surgical intervention the subject walked with flexed knees and a “toe-walking” gait pattern explaining the low values of maximum propulsion force and second maximum of the vertical force during the propulsion phase as well as a very high first maximum vertical force . From the comparison of the results pre and post intervention it can be seen that there is an improvement post surgery, with a diminished first maximum vertical force suggesting better stability and control of the subject as well as an improved maximum propulsion force and improved ratios between the two sides in landing stability, propulsion effort ratio and performance imbalance index.

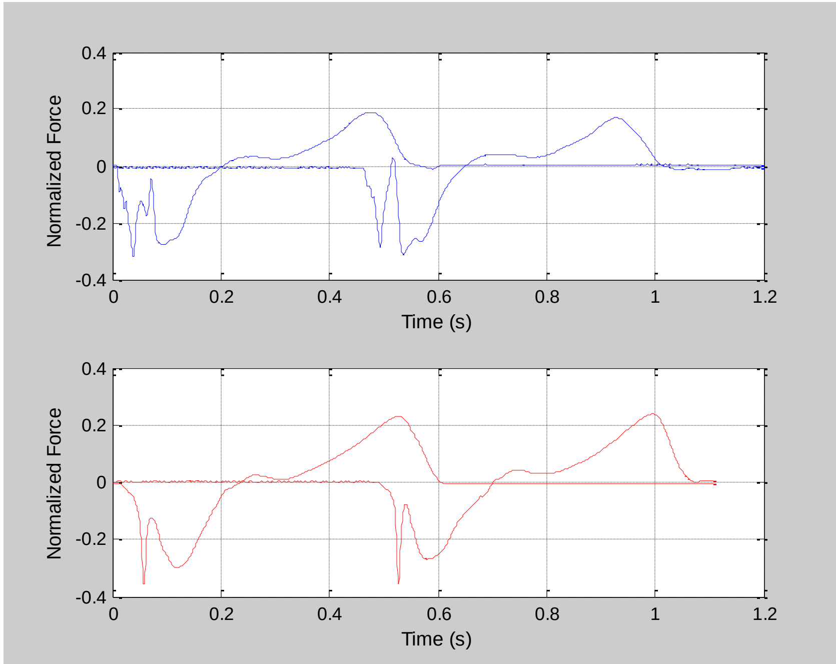

Figure 4.10: GRFs in the A-P direction pre (top figure) and post (bottom figure) intervention for case study one. The first force in both figures is of the left leg and the second of the right.

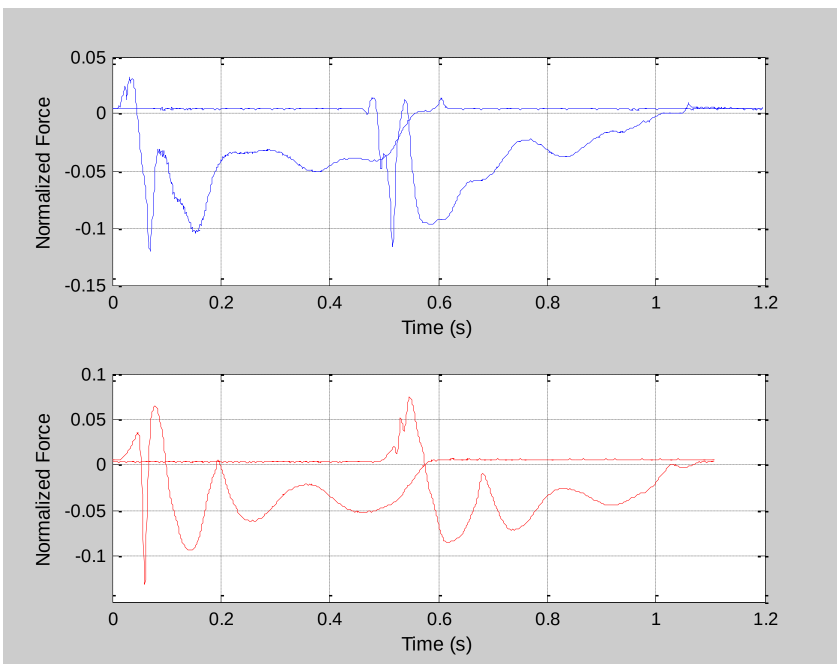

Figure 4.11: GRFs in the M-L direction pre (top figure) and post (bottom figure) intervention for case study one. The first force in both figures is of the left leg and the second of the right.

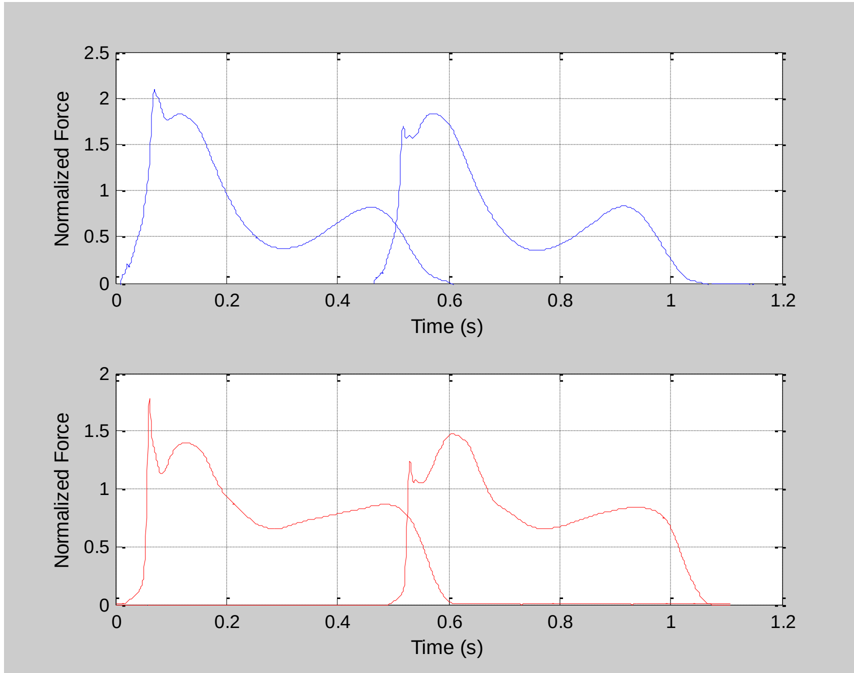

Figure 4.12: GRFs in the vertical direction pre (top figure) and post (bottom figure) intervention for case study one. The first force in both figures is of the left leg and the second of the right.

Table 4.10: Quantitative results for case study one.

	Pre surgery	Post surgery
Maximum braking force [%BW]	-26.59 ± 3.98	-28.82 ± 1.18
Maximum propulsion force [%BW]	16.19 ± 1.62	22.10 ± 1.69
First Maximum of the vertical force [%BW]	170.56 ±16.91	146.23 ± 6.35
Second Maximum of the vertical force [%BW]	85.42 ± 9.03	84.11 ± 1.81
Minimum Vertical Force [%BW]	44.35 ± 9.74	66.32 ± 5.04
Maximum Lateral Force [%BW]	12.24 ± 2.76	10.50 ± 2.96

Table 4.10: Quantitative results for case study one (continued).

	Pre surgery	Post surgery
Maximum Medial Force [%BW]	3.61 ± 2.28	6.55 ± 2.00
Braking Impulse	-0.029 ± 0.003	-0.033 ± 0.002
Propulsion Impulse	0.027 ± 0.003	0.034 ± 0.003
Vertical Impulse	0.51 ± 0.05	0.47 ± 0.02
Lateral Impulse	0.023 ± 0.008	0.023 ± 0.007
Medial Impulse	0.0017 ± 0.0016	0.0020 ± 0.0008
Braking/ propulsion time	0.47 ± 0.10	0.91 ± 0.21
Landing stability ratio	0.84 ± 0.05	1.00 ± 0.05
Propulsion effort ratio	1.04 ± 0.15	1.02 ± 0.10
Performance imbalance index	0.93 ± 0.08	1.00 ± 0.08

4.3.2. Case study two

This case study is of a male with left hemiplegic CP who was seen for a gait analysis 8 months pre (age 12) and 11 months post (age 13) a left sided Jones procedure to correct a clawed hallux, gastrocnemius lengthening, tibialis posterior fractional lengthening, and medial hamstring lengthening. A comparison of post intervention results to pre surgery results shows mild improvement in the force values post surgery. The results are presented in Figures 4.13- 4.15 and Table 4.11. Prior to surgery no significant problems were seen in the subject's walking videos except a mild drop foot gait pattern.

Vertical force values prior to the surgery were low for both legs and there is an improvement post surgery suggesting better control during landing and push off which is also backed up by the landing stability value and maximum braking force values.

Table 4.11: Quantitative results for case study two.

	Non affected side pre surgery	***Affected side pre surgery***	***Non affected side post surgery***	***Affected side post surgery***
Maximum braking force [%BW]	-16.50 ± 1.93	-17.76 ± 3.00	-18.41 ± 2.62	-18.97 ± 1.26
Maximum propulsion force [%BW]	18.09 ± 0.57	14.98 ± 1.02	19.57 ± 1.82	16.22 ± 0.64
First Maximum of the vertical force [%BW]	97.30 ± 2.46	100.46 ± 2.11	115.57 ± 4.69	110.32 ± 3.75
Second Maximum of the vertical force [%BW]	79.32 ± 9.78	76.52 ± 0.77	105.50 ± 4.10	92.53 ± 3.23
Minimum Vertical Force [%BW]	56.16 ± 0.86	51.30 ± 1.70	77.97 ± 2.87	75.79 ± 2.55
Maximum Lateral Force [%BW]	7.25 ± 0.73	12.28 ± 0.69	7.64 ± 1.46	10.37 ± 0.58
Maximum Medial Force [%BW]	2.90 ± 0.49	5.63 ±1.02	4.80 ± 1.60	7.52 ± 1.31
Braking Impulse	-0.027 ± 0.004	-0.019 ± 0.003	-0.035 ± 0.002	-0.029 ±0.004
Propulsion Impulse	0.023 ± 0.001	0.026 ± 0.002	0.033 ± 0.003	0.029 ± 0.004
Vertical Impulse	0.44 ± 0.03	0.39 ± 0.01	0.74 ± 0.04	0.65 ± 0.02
Lateral Impulse	0.027 ± 0.002	0.032 ± 0.002	0.028 ± 0.005	0.034 ± 0.002
Medial Impulse	0.001 ± 0.0002	0.001 ± 0.0003	0.0018 ± 0.0009	0.0026 ± 0.0002
Braking/propulsion time	1.25 ± 0.19	0.87 ± 0.38	1.18 ± 0.08	0.72 ± 0.07

Table 4.11: Quantitative results for case study two (continued).

	Non affected side pre surgery	***Affected side pre surgery***	***Non affected side post surgery***	***Affected side post surgery***
Landing stability ratio	0.76 ± 0.21		0.84 ± 0.12	
Propulsion effort ratio	1.16 ± 0.10		0.90 ± 0.18	
Performance imbalance index	0.94 ± 0.11		0.90 ± 0.18	

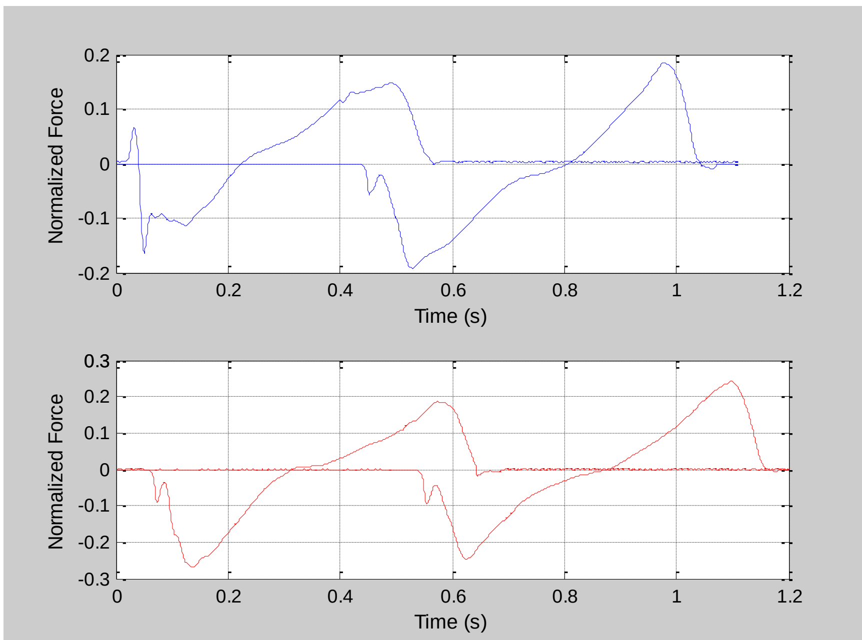

Figure 4.13: GRFs in the A-P direction pre (top figure) and post (bottom figure) intervention for case study two. The first force in both figures is of the left leg and the second of the right.

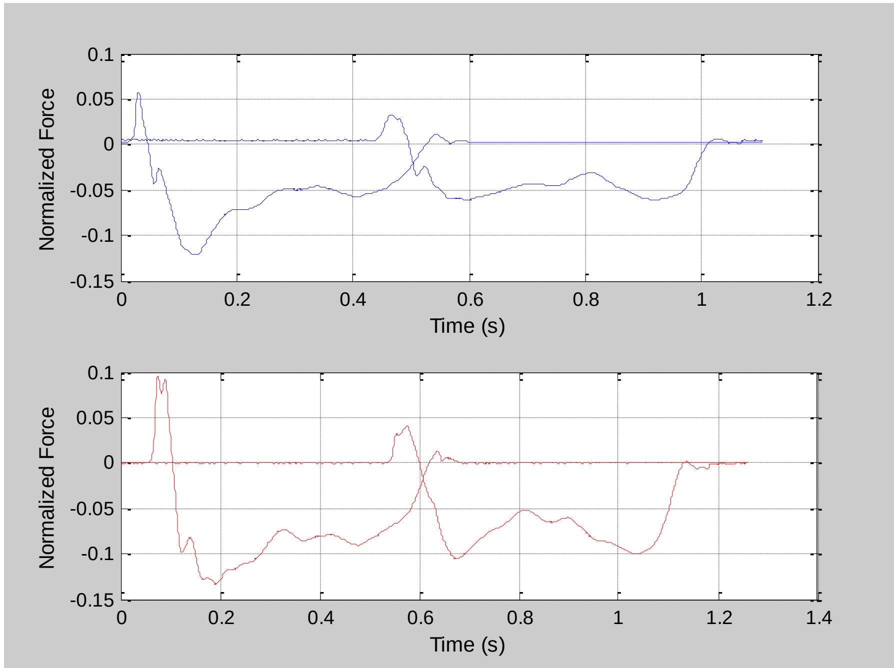

Figure 4.14: GRFs in the M-L direction pre (top figure) and post (bottom figure) intervention for case study two. The first force in both figures is of the left leg and the second of the right.

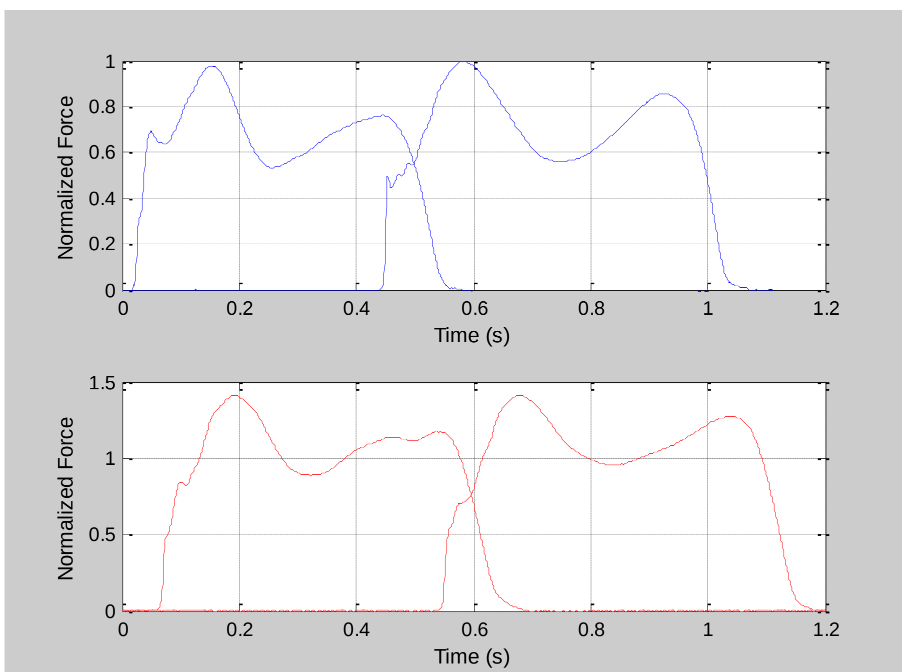

Figure 4.15: GRFs in the vertical direction pre (top figure) and post (bottom figure) intervention for case study two. The first force in both figures is of the left leg and the second of the right.

4.3.3. Case study three

The third case study is of a 14 year old female with diplegic CP who was seen for a gait analysis with and without bilateral ankle-foot orthoses. The comparison of braces vs. barefoot walking is presented in figures 4.16- 4.18 and in Table 4.12. Very few improvements are seen with the subject walking in her braces. The braces seem to affect the forces exerted during gait and the subject has more variability with the braces on. As a result of lower force values the propulsion impulse is reduced as well.

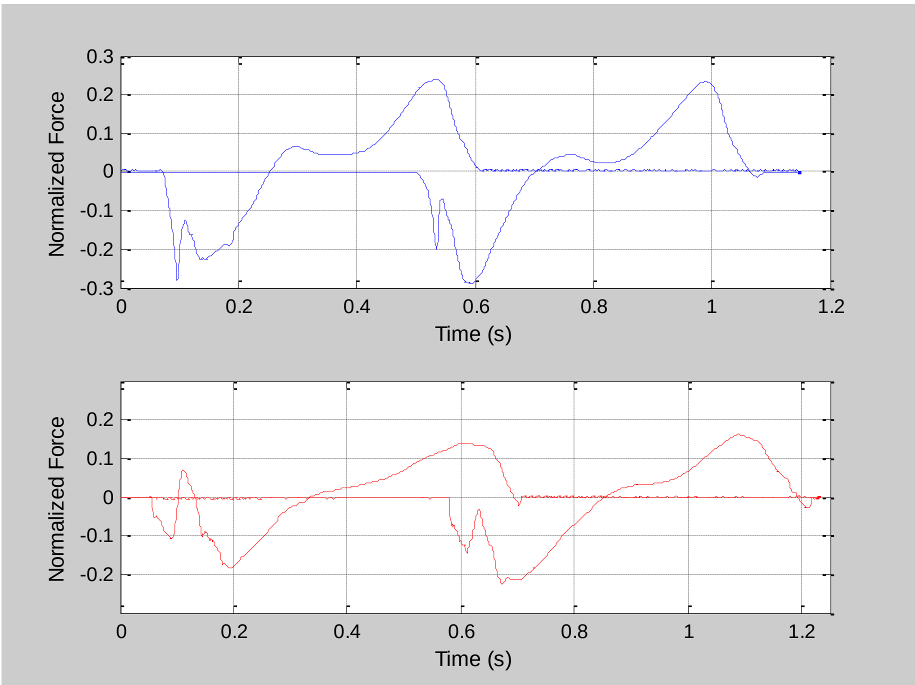

Figure 4.16: GRFs in the A-P direction without (top figure) and with (bottom figure) braces for case study three. The first force in both figures is of the left leg and the second of the right.

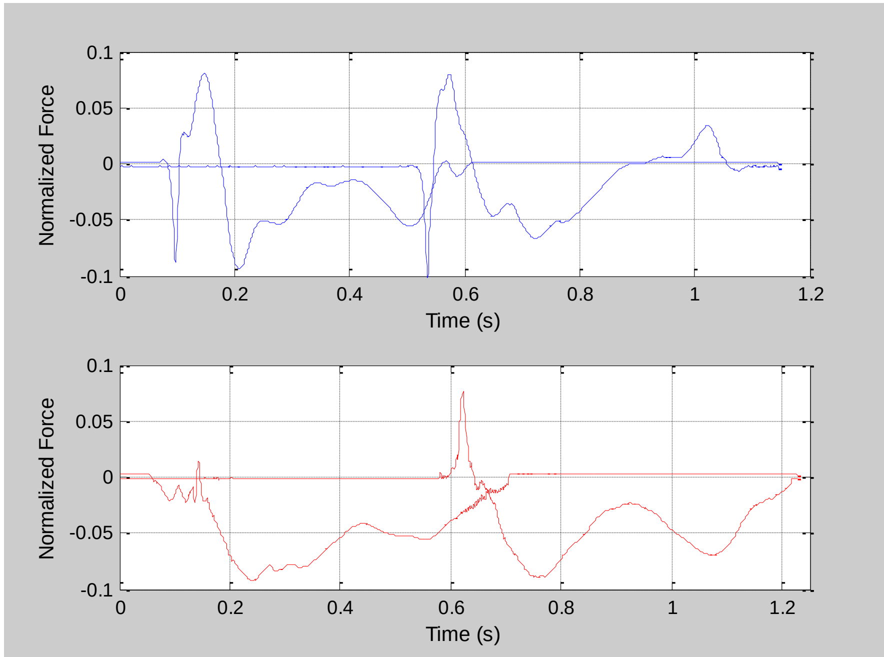

Figure 4.17: GRFs in the M-L direction without (top figure) and with (bottom figure) braces for case study three . The first force in both figures is of the left leg and the second of the right.

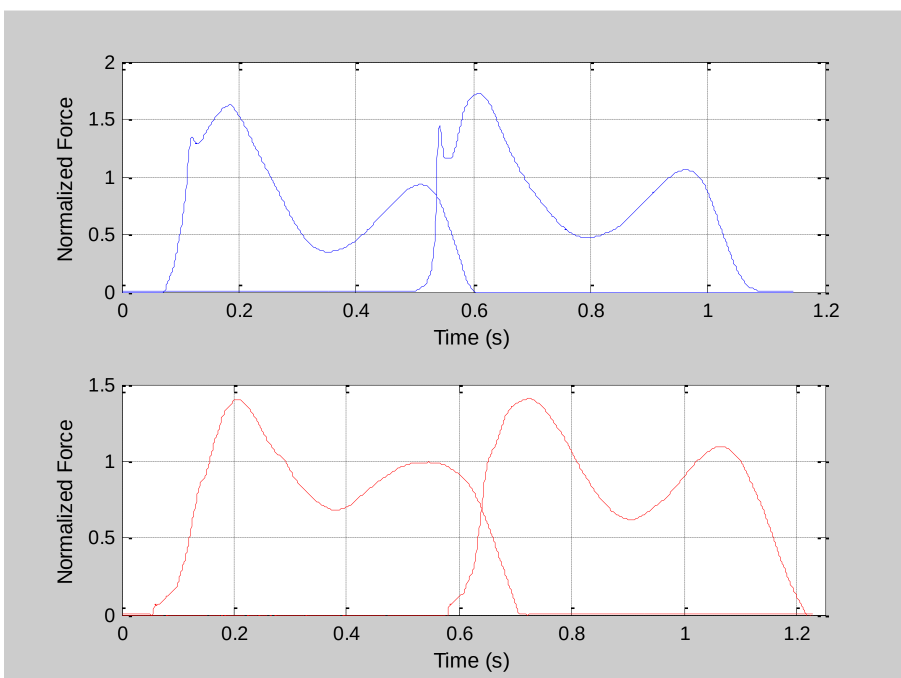

Figure 4.18: GRFs in the vertical direction without (top figure) and with (bottom figure) braces for case study three. The first force in both figures is of the left leg and the second of the right.

Table 4.12: Quantitative results for case study three.

	Barefoot	***Braces***
Maximum braking force [%BW]	-28.06 ± 2.85	-21.43 ± 3.08
Maximum propulsion force [%BW]	21.60 ± 2.27	14.41 ± 1.29
First Maximum of the vertical force [%BW]	161.09 ± 6.46	137.76 ± 2.42
Second Maximum of the vertical force [%BW]	99.17 ± 6.73	99.90 ± 6.83
Minimum Vertical Force [%BW]	49.16 ± 3.10	67.68 ± 0.74
Maximum Lateral Force [%BW]	8.66 ± 2.77	8.96 ± 0.1
Maximum Medial Force [%BW]	7.33 ± 3.33	5.18 ± 4.67
Braking Impulse	-0.031 ± 0.005	-0.027 ± 0.006
Propulsion Impulse	0.032 ± 0.002	0.025 ± 0.002
Vertical Impulse	0.50 ± 0.03	0.52 ± 0.004
Lateral Impulse	0.013 ± 0.007	0.029 ± 0.0025
Medial Impulse	0.005 ± 0.003	0.001 ± 0.0008
Braking/ propulsion time	0.49 ± 0.04	0.75 ± 0.02
Landing stability ratio	0.84 ± 0.18	1.11 ± 0.62
Propulsion effort ratio	0.89 ± 0.06	0.94 ± 0.04
Performance imbalance index	0.86 ± 0.10	1.22 ± 0.03

4.4. Results for the energy expenditure as measured by the Actiheart device (Aim 5)

An Actiheart device was placed on all subjects in the prospective data collection and used to measure the energy expenditure during ambulation. The results from the device's software are presented in Figure 4.19. The results were also averaged to examine the difference between the populations. Since only two subjects had hemiplegic CP and five had diplegic CP, comparison between typically developed children and children with CP were calculated (Table 4.13).

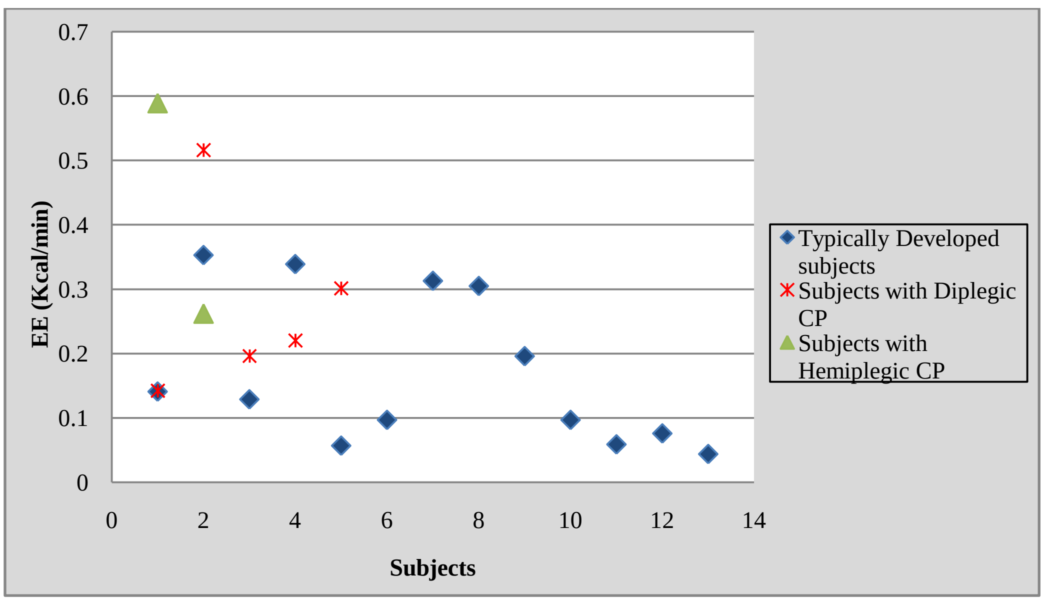

Figure 4.19: EE results for all subjects who participated in the prospective data collection.

Table 4.13: Averaged values of EE for typically developed children and children with CP.

	EE value (Kcal/min)
Typically developed children (N=13)	0.169 ± 0.119
Children with CP (N=7)	0.318 ± 0.169

4.4.1. Statistical results for the analysis of relationship between force values and EE

Correlation and ANOVA analysis were run on the available data to determine whether there is a consistent connection between different force parameters and the EE values calculated from the Actiheart. The results are presented in Table 4.14.

Table 4.14: Significant results for the correlation analysis between EE and the 17 parameters derived from GRFs.

	Correlated to:
Typically developed children (N=12)	Maximum braking force (r = 0.66)
Children with CP (N=7)	Maximum braking force (r = 0.65) Minimum vertical force (r = -0.50)

4.5. Results for GRF characteristics derived from accelerometer data (Aim 4)

Eight characteristic parameters of the vertical and A-P GRF curves were derived from acceleration data and compared to the parameters calculated from the GRF data. These parameters included the braking, propulsion and vertical impulses, the peak braking and propulsion forces and the two peaks and minimum between them of the vertical force. Since the parameters were calculated using several methods the results were statistically analyzed to evaluate and compare the different methods.

The motion lab in which data collection took place has four force plates and therefore data from 4 different force plates were available. In the first step of the analysis a factorial ANOVA was run to examine the main effects of the analysis. It was discovered that there were main effects of force plates; with two of the force plates showing different patterns of values. These two force plates were the two end force plates which may be impacted by gait initiation and termination by the subjects, therefore their results were taken out of the analysis. A Pearson correlation test between the force data and the accelerometers results was run on the remaining force plate data.

The result for the correlation test in typically developed children walking at their chosen walking velocity is presented in Figure 4.20. Quantitative tabular results for the statistical analysis presented in this section can be found in Appendix C.

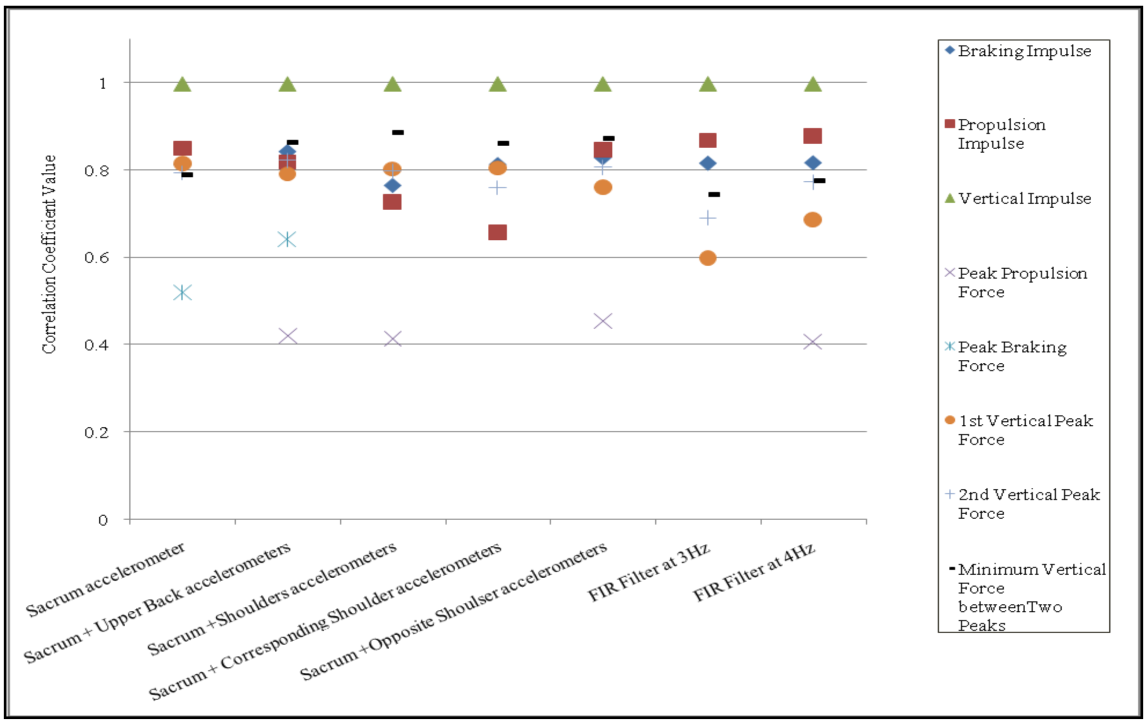

Figure 4.20: Results for the correlation test between force plate data and the values calculated from acceleration data for typically developed children walking at their self selected walking speed.

Notes for Figures 4.20- 4.27:

1. The methods presented in the figures, from left to right are: sacrum accelerometer, sacrum and upper back accelerometer, sacrum and both shoulders accelerometers, sacrum and the corresponding shoulder accelerometers, sacrum and the opposite shoulder accelerometers, sacrum accelerometer with a FIR filter at 3Hz and sacrum accelerometer with a FIR filter at 4Hz.

In the next analysis step the data of children with CP were added. The correlation test was run again and the results can be seen in Figure 4.21.

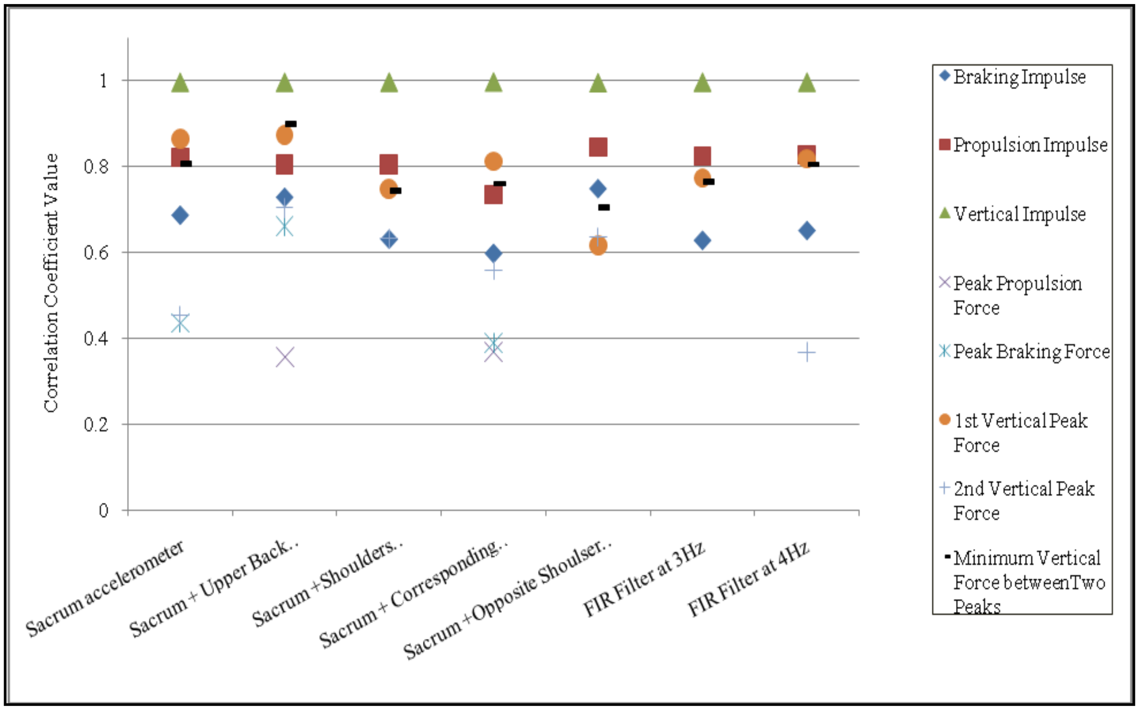

Figure 4.21: Results for the correlation test between force plate data and the values calculated from acceleration data for typically developed children and children with CP walking at their self selected walking speed.

The same analysis was performed on the two additional conditions, in which the subjects walked faster and slower than their chosen walking velocity. These analyses can be seen in Figures 4.22-4.27. For the fast walking condition the factorial ANOVA showed the same results as in the chosen walking velocity condition and the analysis included only the two middle force plates. As for the slow velocity trials, the results were different.

In this condition it was found the side, left or right leg, had an effect therefore the first and third force plates had similar results, and the second and fourth force plates also had similar results.. The analysis of the slow velocity trials was run twice, for the grouping of force plates as stated above, and is presented for first and third force plates and second and fourth force plates separately.

To summarize the results for the different methods and three walking speeds the correlation results for the eight ground reaction force parameters, for each method, were averaged. For cases in which there was no correlation present a value of zero was given and added to the average calculation. The comparison of the average correlation values for the three walking speeds and different methods can be seen in Tables 4.15-4.16.

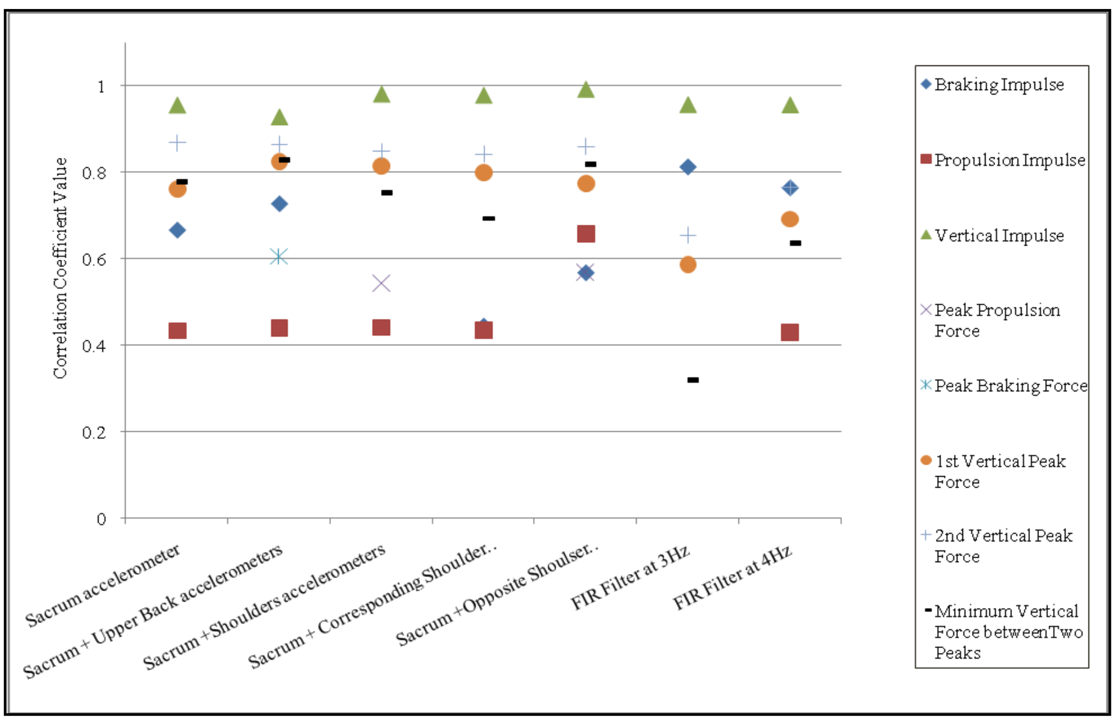

Figure 4.22: Results for the correlation test between force plate data and the values calculated from acceleration data for typically developed children walking faster than their self selected walking speed.

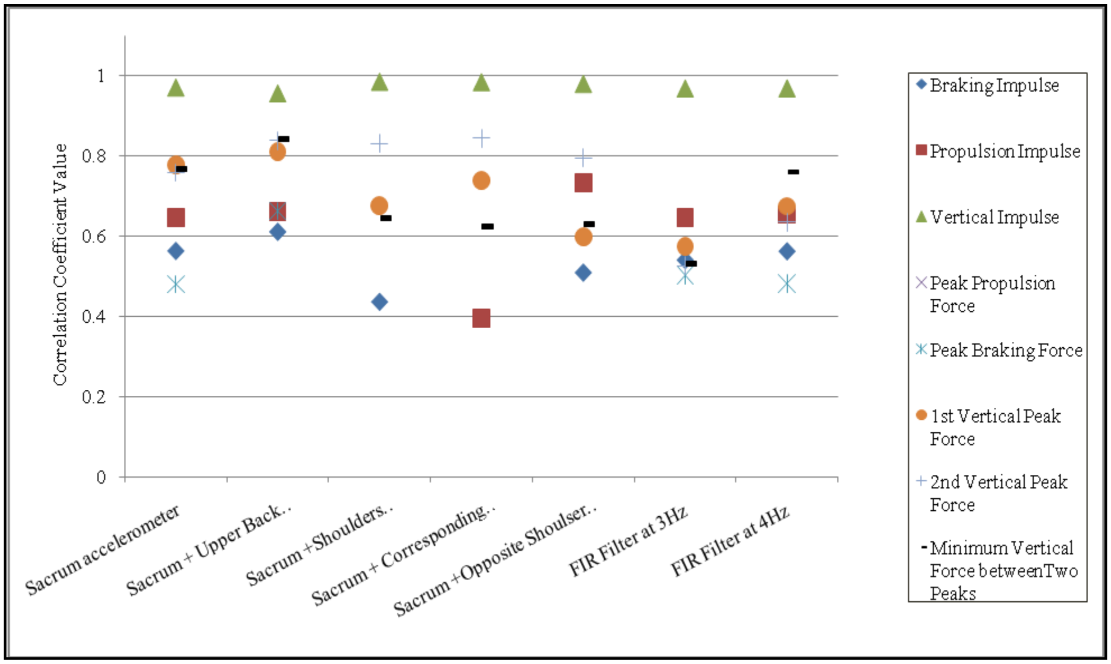

Figure 4.23: Results for the correlation test between force plate data and the values calculated from acceleration data for typically developed children and children with CP walking faster than their self selected walking speed.

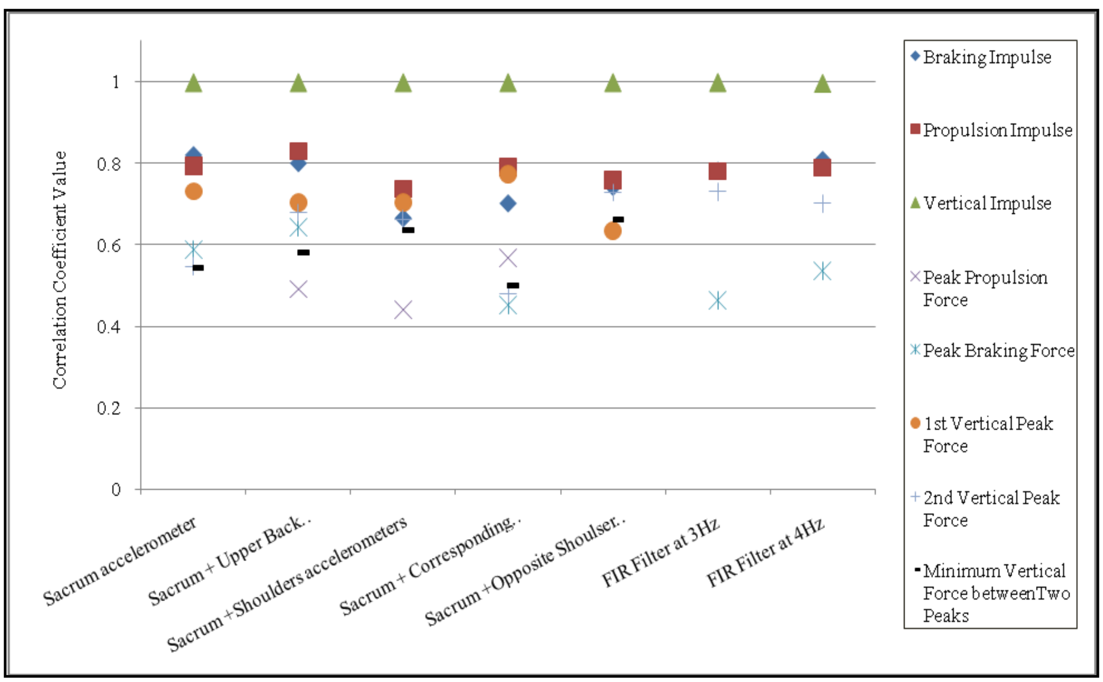

Figure 4.24: Results for the correlation test between force plate data and the values calculated from acceleration data for typically developed children walking slower than their self selected walking speed on the first and third force plate.

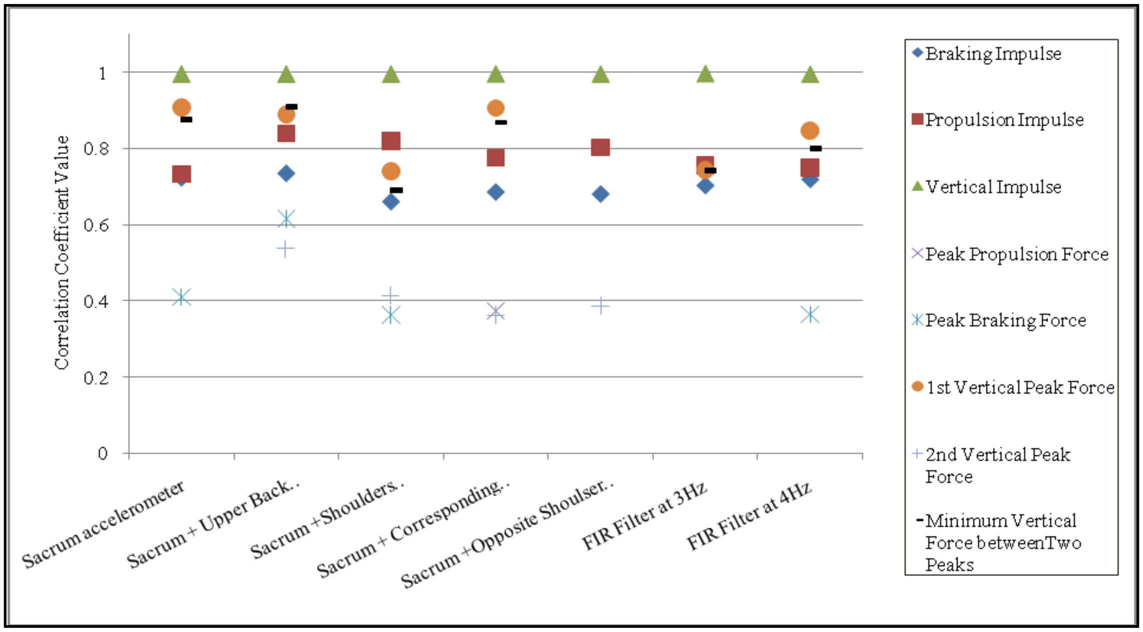

Figure 4.25: Results for the correlation test between force plate data and the values calculated from acceleration data for typically developed children and children with CP walking slower than their self selected walking speed on the first and third force plates.

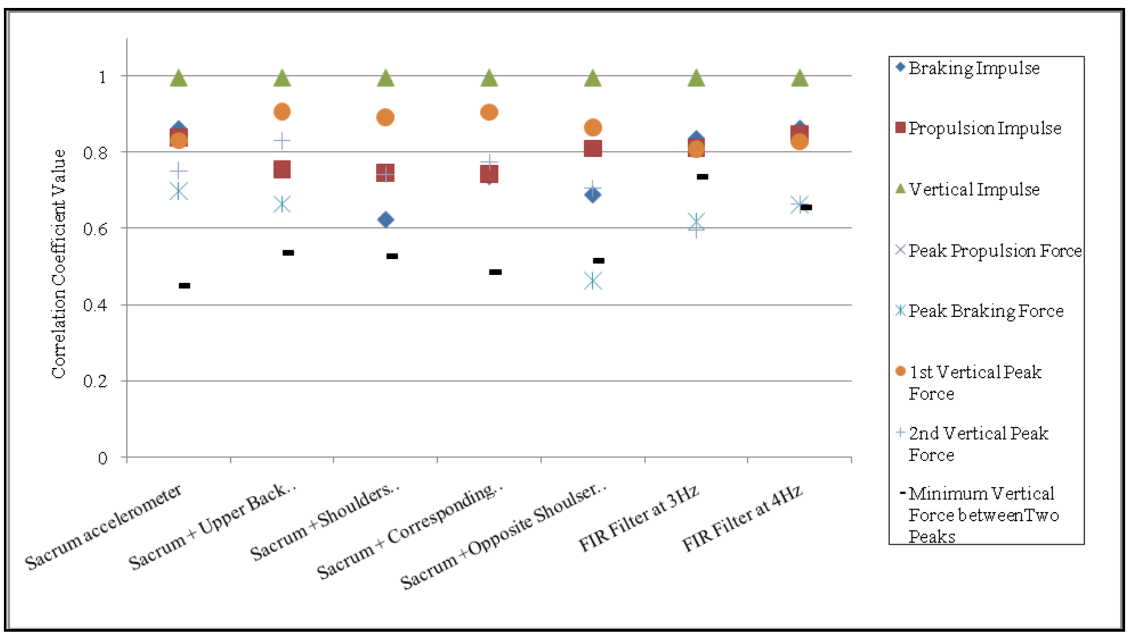

Figure 4.26: Results for the correlation test between force plate data and the values calculated from acceleration data for typically developed children walking slower than their self selected walking speed on the second and fourth force plates.

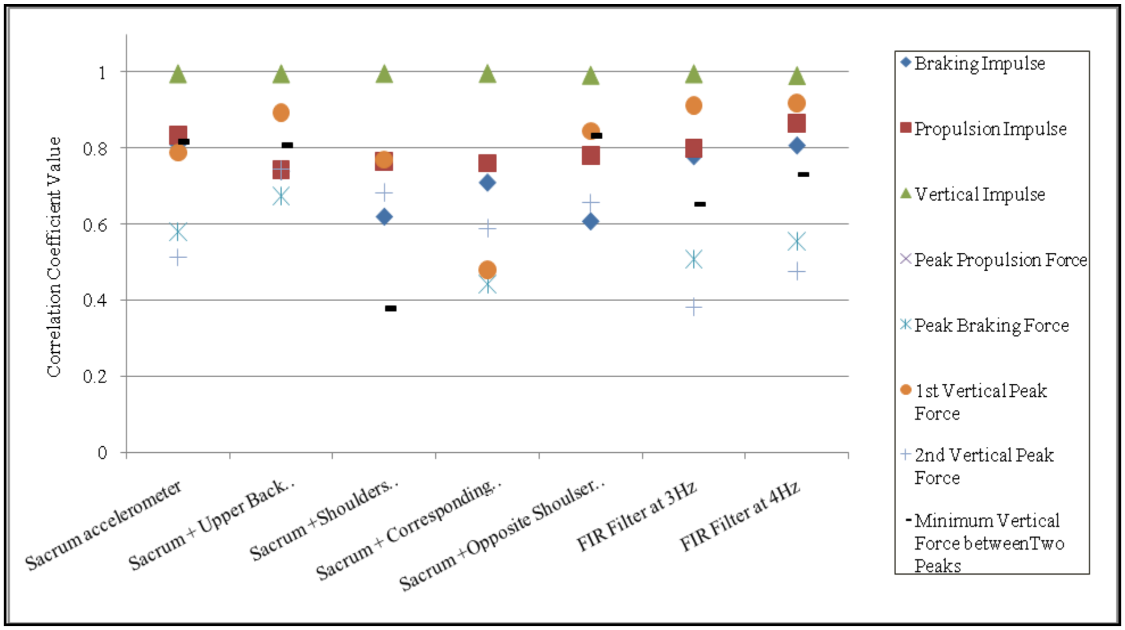

Figure 4.27: Results for the correlation test between force plate data and the values calculated from acceleration data for typically developed children and children with CP walking slower than their self selected walking speed on the second and fourth force plates.

Table 4.15: Results of the average correlation value for the different methods used to calculate GRFs characteristics from acceleration data for self selected and fast walking speeds.

	TD-self selected walking speed	***TD and CP-self selected walking speed***	***TD-faster than self selected walking speed***	***TD and CP-faster than self selected walking speed***
Sacrum accelerometer	0.70 ± 0.31	0.63 ± 0.32	0.56 ± 0.38	0.62 ± 0.29
Sacrum and upper back accelerometers	0.78 ± 0.17	0.75 ± 0.19	0.65 ± 0.31	0.67 ± 0.30
Sacrum and shoulders accelerometers	0.67 ± 0.32	0.57 ± 0.37	0.55 ± 0.38	0.51 ± 0.39
Sacrum and corresponding shoulder accelerometers	0.61 ± 0.39	0.65 ± 0.22	0.52 ± 0.37	0.45 ± 0.41
Sacrum and opposite shoulder accelerometers	0.70 ± 0.32	0.57 ± 0.37	0.66 ± 0.30	0.53 ± 0.36
FIR filter at 3Hz	0.59 ± 0.38	0.57 ± 0.40	0.38 ± 0.42	0.54 ± 0.26
FIR filter at 4Hz	0.67 ± 0.32	0.56 ± 0.39	0.53 ± 0.36	0.59 ± 0.28

Tables 4.15-4.16 Notes:

1. "TD" stands for typically developed children.
2. The two highest correlation values for each walking condition are highlighted.

Table 4.16: Results of the average correlation value for the different methods used to calculate GRFs characteristics from acceleration data for the slow walking speed.

	TD-slower than self selected walking speed-first and third force plates	***TD and CP-slower than self selected walking speed-first and third force plates***	***TD-slower than self selected walking speed-second and fourth force plates***	***TD and CP-slower than self selected walking speed-second and fourth force plates***
Sacrum accelerometer	0.63 ± 0.30	0.66 ± 0.35	0.68 ± 0.32	0.68 ± 0.31
Sacrum and upper back accelerometers	0.72 ± 0.16	0.69 ± 0.32	0.68 ± 0.31	0.70 ± 0.30
Sacrum and shoulders accelerometers	0.61 ± 0.29	0.59 ± 0.31	0.57 ± 0.38	0.53 ± 0.37
Sacrum and corresponding shoulder accelerometers	0.66 ± 0.19	0.62 ± 0.34	0.58 ± 0.39	0.50 ± 0.35
Sacrum and opposite shoulder accelerometers	0.57 ± 0.37	0.36 ± 0.42	0.63 ± 0.31	0.59 ± 0.38
FIR filter at 3Hz	0.47 ± 0.41	0.49 ± 0.42	0.68 ± 0.30	0.63 ± 0.32
FIR filter at 4Hz	0.48 ± 0.42	0.56 ± 0.39	0.69 ± 0.30	0.67 ± 0.32

Chapter 5: Discussion

Gait analysis is a common tool for the assessment of ambulation in normal and pathological populations. In order to ensure the GRFs values are reproducible and accurate, the first step is to define whether the gait cycles within a given trial or walk over sequential force plates are consistent as measured by a consistency test.

5.1. Consistency of gait in typical and pathological children's populations (Aim1)

A measure of GRF consistency has been reported and validated in adults (Seliktar, 1979). This works expanded upon this previous work by application of the consistency measure to typically developed children and children with CP. This measure provided the range of values in which a child's gait can be defined as consistent. This test measured the deviation from a completely consistent gait cycle, in which the sum of braking and propulsion forces of the two legs amount to zero, and suggest a consistent gait in children is one in which the deviation is not greater than 6.5%. The consistency test results were applied to all other analyses involving GRFs; therefore, only consistent gait cycles were taken for further analyses.

5.2. Use of GRF data in the analysis of gait (Aim 2)

5.2.1. Normative GRF data for children

Although GRF data are an integral part of gait analysis, no reported studies have systematically quantified the differences in GRF data between groups of typically developed children and children with CP.

Engsberg et al. (Engsberg, 1993) reported normative GRF data for able bodied and below knee children amputees. Their results provide evidence of concurrent validity for the results of this study. The comparison between the values of typically developed children in the two studies (Table 5.1) shows the majority of parameters in both sets of results are within 11 percent of each other and the larger differences noted could be a result of either the difference in the sample sizes or the use of the consistency test to ensure quality data in the current study. The majority of parameters from the current study also have a smaller standard deviation value. Even though the sample size in Engsberg's study was much larger, which should decrease the standard deviation, the application of the consistency test prior to data analysis was helpful in reducing the standard deviation. The most significant difference between the two studies is of the value of the normalized medial-lateral impulse which is consistent with the prior knowledge of the highly variable nature of this parameter. In their findings, Engsberg et al. also found no significant differences between the right and left leg. Therefore in our analysis, we only differentiated between legs in children with hemiplegic CP which have clear clinical differences in the legs.

Table 5.1: A comparison of the GRF data received in the study done by Engsberg et al and the current study presented in this work.

	Engsberg et al. (N=450)	***Current Study (N=54)***
Maximum braking force [%BW]	-24 ± 6	-18.99 ± 4.77
Maximum propulsion force [%BW]	24 ± 5	22.08 ± 3.50
First Maximum of the vertical force [%BW]	128 ± 17	114.74 ± 10.81
Second Maximum of the vertical force [%BW]	119 ± 13	110.92 ± 7.52
Minimum vertical force between the two peaks [%BW]	63 ± 13	76.66 ± 8.11
Maximum Medial force [% BW]	7 ± 5	5.17 ± 2.26
Maximum Lateral force [% BW]	8 ± 5	5.82 ± 1.73
Normalized Braking Impulse [BW*T]	-0.028 ± 0.006	-0.0277 ± 0.0051
Normalized Propulsion Impulse [BW*T]	0.030 ± 0.006	0.0282 ± 0.0049
Normalized Vertical Impulse [BW*T]	0.45 ± 0.05	0.50 ± 0.04
Normalized Medial Impulse [BW*T]	0.007 ± 0.009	0.0027 ± 0.0015
Normalized Lateral Impulse [BW*T]	0.014 ± 0.01	0.014 ± 0.007

5.2.2. Comparison of GRF data in typically developed children and children with CP

The analysis of the GRF data, in typically developed and CP populations, produced interesting results and provides an alternative, quantitative, way to assess gait deviations as well as changes due to time or interventions.

Results for the A-P component of the GRF showed that in two groups (typically developed children and the non affected side of children with hemiplegic CP) the maximum braking force was smaller in absolute value when compared to the propulsion force. The opposite trend was found in children with diplegic CP and the affected side of children with hemiplegic CP. This difference in the maximum braking and propulsion forces between children with diplegic CP and typically developed children was statistically significant. Although the same trend was found in the affected side of children with hemiplegic CP, a significant difference between this group and typically developed children was found only for the maximum propulsion force suggesting the braking peak forces in children with hemiplegic CP remain within the range found in typically developed children and the pathological condition is stronger in the propulsion phase with higher peak propulsion forces produced. These findings suggest that in children with diplegic CP, impairments such as drop foot or weakness of the muscles diminish their control during ambulation with greater utilization of braking rather than propulsive forces during level gait.

The results for the comparison of brake phase time and propulsion phase time did not show conclusive results. In the clinical populations of diplegic and affected side in hemiplegic CP, the brake time was smaller than the propulsion time; however, the variability was high. Despite this, the statistical analysis revealed a significant difference between both of these groups and typically developed children with children with CP spending more time pushing forward than typically developed children. Combining the two comparisons mentioned above, it can be concluded that children with CP spend more time in the propulsion phase rather than generate the peak propulsion forces found in typically developed children.

Examination of the peak forces in the vertical direction showed that in the typical population the two peaks were close in value and the second peak was slightly lower in value. The same results followed for the other populations but it was the difference in values between the two peaks that stood out. The highest difference between the peak forces was found in children with diplegic CP. Children with hemiplegic CP showed a higher difference between peak vertical forces on the affected side but not as great a difference as in children with diplegic CP. The minimum force between the two peaks was in the range of 70% for all four populations. The interesting finding was that in children with hemiplegic CP, data from the non affected leg showed a trend towards the lowest force values when compared to typically developed children and even children with diplegic CP. This suggests that a compensation mechanism is used in this group in which the non affected leg produces a higher second peak force, during push off, to compensate for the affected leg with greater compensatory knee flexion in midstance. The statistical analysis results showed a significant difference between typically developed children and children with diplegic CP in values of the first vertical peak, and a significant difference between typically developed children and children with diplegic and hemiplegic CP (the affected side) for the second vertical peak. There was no statistical significant difference among the populations for the minimum vertical force between the two peak forces, suggesting this parameter may not be sensitive enough to detect related gait deviations in children with CP.

In the medio-lateral direction, some differences were seen, noticeably in children with diplegic CP, but the high variability of forces in this plane suggests caution in any conclusion made. In children with diplegic CP higher lateral peak forces were present which could be the result of in toe walking or a scissoring gait, which are common gait abnormalities seen in this population, causing more medially directed foot forces.

The normalized impulses of GRFs in the three directions were also examined. The braking and propulsion impulses were examined in two different ways. The values of the normalized impulses are presented in Table A4. The expected results would be ones in which the difference between the braking and propulsion impulse values is close to zero indicating the leg puts the same amount of effort to both phases of the gait cycle. The results for the amount of deviation from zero (Table 5.2) showed that in typically developed children and children with diplegic CP the difference between the two impulses is less than 1%. Children with hemiplegic CP exhibited greater asymmetry as indicated by the higher differences noted on both affected and non-affected sides. The results show that in the non affected side the propulsion impulse is higher than the braking impulse and in the affected side the opposite happens, which suggests the non affected side compensates for the affected side during push off which in turn results in a larger braking impulse on the affected side. The ANOVA analysis showed a significant difference between the affected side of children with hemiplegic CP both for braking and propulsion impulse but did not show any significant difference between values from the non affected side and typically developed children. This result suggests that the non-affected leg uses relatively typical mechanics, and the compensation is mapped on to this relatively typical pattern.

Table 5.2: The difference between propulsion and braking impulse presented as percentage of the total value of the impulse.

	Difference between the normalized propulsion and braking impulse
Typically Developed Children	0.9%
Children with Diplegic CP	0.2%
Children with Hemiplegic CP-non affected side	2.9%
Children with Hemiplegic CP-affected side	-3.4%

Table 5.2 Notes:

1. A negative value is a result of the braking impulse being larger in value the propulsion impulse and a positive value is the result of the propulsion impulse value being larger than the braking impulse value.

The results for the calculations of propulsion to braking impulse ratio (Figure 4.2, Table 4.6 and Table A2) were not as statistically significant as the non ratio values themselves. This suggests that looking at the impulse values rather than the ratio provide better, more informative results.

The normalized impulse value in the vertical direction (Table 4.5) was found to have an average value of 0.5 and is most useful in providing another tool to assess the asymmetry between the body’s two lower limbs and especially the difference between the affected and non affected sides in children with hemiplegic CP. Statistical analysis also showed a significance difference is these values between both populations of children with hemiplegic CP and typically developed children pointing again to the compensation mechanism that develops in the non affected side of children with hemiplegic CP.

In the M-L direction both the lateral and medial impulses were examined. The medial impulse has small values and showed large variability in all four populations. The lateral impulse values were an order larger than the medial impulse values as would be expected from the pattern of the force curves. Children with diplegic CP had the largest values of lateral impulse and statistical analysis found this value, in children with diplegic CP, had a significant difference when compared to typically developed children suggesting again that in toe walking and scissor gait found in this population could be the reason for the large values. It is important to remember that the strength of the M-L force is in the indication of whether stability or the lack of it exists and that is best seen in the curve patterns of the M-L force.

The within subject group correlation analysis of the fourteen parameters (Table 4.7) provided useful information as well. The correlation between the peak propulsion force and the propulsion impulse showed good to excellent correlations in typically developed children and both sides of children with hemiplegic CP. This finding implies that higher impulses are produced by higher values of forces, rather than by increasing the time of application. In children with diplegic CP, the correlation was only a moderate one which would imply that they achieved higher values of propulsion impulses, by increasing the duration of the propulsion phase. Similar results were seen for the peak braking force and braking impulse as well, with the exception of the affected side of children with CP which had a lower correlation value. Another result that supports the compensation mechanism found between the two legs in children with hemiplegic CP was seen in the correlation value for braking and propulsion impulses. The correlation values were good to excellent for typically developed children and children with diplegic CP but not for both sides of children with hemiplegic CP. The correlations between peak propulsion force and braking impulse, and peak braking force and propulsion impulse revealed similar results for each population reflecting the continuity across gait cycles. A higher propulsion force would lead to a bigger braking impulse and a higher braking force will cause a higher propulsion impulse.

The correlations between the four vertical components: vertical impulse, two vertical peaks and the minimum force between them, showed the only relationships and linear connections were between the minimum vertical force between the two peaks and the two vertical peaks. In typically developed children and for both sides of children with hemiplegic CP, a higher first vertical peak was correlated to a lower minimum vertical force. Another correlation found in typically developed children and the non affected side of children with hemiplegic CP was a lower minimum vertical force corresponded to a higher second vertical peak force. However, in children with diplegic CP the opposite relationship existed which implies their muscles are not strong enough to produce high push off forces once their knees are flexed in midstance.

The correlation results also demonstrate linear relationships exist between the different forces components in different planes. The peak propulsion force had low correlation to the two vertical peak forces and minimum vertical force between them in typically developed children and the non affected side of children with CP. One exception to this was the difference for the maximum propulsion force and minimum vertical force (children with CP had only partial correlations for these 3 cases). In typically developed children and the affected side of children with hemiplegic CP, this correlation indicated a lower minimum vertical force would produce a higher propulsion peak force. However for children with diplegic CP and the non affected side of children with CP the opposite relationship was found.

The peak braking force showed significant correlation with the first vertical peak in typically developed children and the non affected side of children with CP. This indicates a higher level of motor control exists in these groups compared with the pathological cases. In the affected side of children with hemiplegic CP the correlation values were reduced and in the diplegic population no relationship between the parameters was found. Two relationships, with moderate to good correlations, between the first vertical peak force and the braking and propulsion impulses were due noted only in typically developed children and the non effected side of children with hemiplegic CP. This suggests that the smoothness of transitions within a gait cycle is more apparent is these populations than in the pathological populations.

5.2.3. The three parameters: landing stability ratio, propulsion effort ratio and performance imbalance index, derived from GRF data

Three new measures were defined to help assess the performance of a subject while walking. These measures were: landing stability ratio, propulsion effort ratio and the performance imbalance index. The measures, calculated from impulse values, looked at the relationship between the two legs during the gait cycle and therefore differ from the other force characteristics, previously reviewed. Since the measures take both legs into account only three subject groupings were present: typically developed children, children with diplegic CP and children with hemiplegic CP. For all three measures the desired value was 1 indicating a balanced effort of both legs during locomotion. The average value of the landing stability ratio was close to one for all three populations and the main difference was in the amount of variability. In the pathological populations the variability was larger than in typically developed children. The propulsion effort ratio and performance imbalance index showed more difference with reduced values for the hemiplegic population where a clear difference in clinical impairments exists between the two legs. The results of the statistical analysis also showed the same trend for children with hemiplegic CP since significant difference between this pathological population and typically developed children was found for the propulsion effort ratio and performance imbalance index. The Pearson correlation test showed moderate correlations of the landing stability to both the propulsion effort ratio and the performance imbalance index in typically developed children and children with hemiplegic CP. The correlation was not so high as to suggest these parameters are measuring the same underlying characteristics. In children with diplegic CP, moderate correlations of the performance imbalance index to both landing stability and propulsion effort ratio were found.

5.3. The use of GRF data for within subject analysis

These approaches using GRF data were applied to three case studies of typical scenarios found in gait analysis. Two of the subjects had undergone surgical interventions and were seen pre and post intervention for a gait analysis, and the third subject was seen for a gait analysis with and without orthotics to assess their contribution. The graphs and quantitative values give an informative picture as to the changes due to the intervention.

In case study one improvement can be seen in all planes of motion. In the vertical direction the difference between the first and second maximum of the vertical force diminishes which is also the result of a smaller first maximum, closer in value to typically developed children. The values themselves are less variable with lower standard deviation values. An improvement is also seen in the values of the vertical minimum force between the two peaks. Although a lower average value than that found in typically developed children was noted, the value still improved from the pre surgery value. In the M-L direction, the maximum medial and lateral forces were similar which is what is found in typically developed children. In the A-P direction, the ratio of braking time to propulsion time improves, increasing to values closer to those of typically developed children. This change in ratio indicated improved push off capabilities also supported by an increase in the peak propulsion force which is in the range of the values seen in typically developed children. The three parameters that look at the relationship between the left and right side also improve after intervention showing improved symmetry between the two sides.

Case study two did not show such clear improvements as seen in case study one. In the vertical direction it was surprising to see the non affected leg had lower first maximum values compared to the affected leg but after intervention that trend switched. An improvement was seen in the minimum vertical force, between the two peaks, with values increasing and getting closer in value to normal values. In the A-P direction there was an increase in the peak propulsion and breaking forces showing the subject generated higher forces closer in values to those found in typically developed children. Higher impulse values were seen in all three planes suggesting improved strength post surgery. The landing stability ratio, propulsion effort ratio, and performance imbalance index also did not show a significant improvement suggesting either the surgery did not have the intended affect or that perhaps the subject was still adjusting to the changes in his movement ability at the time of the gait analysis.

Case study three compared barefoot and braced conditions in a subject with diplegic CP. Significant differences between conditions can be seen in the A-P forces, where the peak forces seen with braces are reduced. The decrease in the peak braking force suggests better control with the brace but it also affects the ability of the subject to push off as seen by the decrease in the peak propulsion force value. In a related finding the ratio of braking phase time to propulsion phase time of the gait cycle increases and shows a better ratio between the two values which suggests the subject spends less time pushing off. Coupled with the decrease in propulsion impulse values the findings demonstrate the subject has less push off capabilities with the braces. Improvements can be seen in the vertical forces in general, the first peak's value is reduced and the minimum value between the two peaks increases bringing these values closer to typical values. In the M-L direction, the values show an improvement in stability with a reduced lateral impulse value with the braces on as evident on the corresponding graphs. All other parameters did not show significant differences when comparing barefoot to braced conditions.

5.4. The relationship between EE and GRFs (Aim 5)

As been previously mentioned, the relationship between mechanical work and energy expenditure has been studied but no one has reported the relationship between characteristics of GRFs and the energy expended during ambulation.

The results of the EE data received from the Actiheart device did not show a clear trend and it must be emphasized that although the Actiheart was shown to be reliable in typically developed children, such validations had not been completed in children with CP. Overall as would be expected, the average value of EE in children with CP was higher than that found in typically developed children. This increase in EE has been attributed to less efficient gait pattern and greater co-activation of muscles. The EE values were tested for the existence of a relationship to the 17 GRF values and measures examined in this work and in both populations the EE value had good correlation values ($r=0.66$ for typically developed children and $r=0.65$ for children with CP) to the maximum braking force. In the groups with CP there was also a moderate correlation ($r=-0.50$) to the minimum vertical force. The latter suggests that the deviations of the minimum vertical force, found in children with CP, for reason such as stiff knee gait as well as the compensation mechanism seen in hemiplegic CP may underlie the increase in EE values. This relationship needs further analysis with a larger sample size. As for the maximum braking force, which may relate to the amount of control a subject has during landing, the values of maximum braking force were found to be higher in the CP population, specifically the diplegic population which was the majority of the CP subjects in this prospective study. Therefore, it suggests a possible connection between the amount of control during landing and heel strike and the amount of energy the subject expends during ambulation.

5.5. Derivation of GRFs characteristics from acceleration data (Aims 3 and 4)

First, the dataset of the eight parameters of GRF data derived from acceleration data, in the vertical and A-P direction was analyzed for effect of force plate using an ANOVA. After the ANOVA test was done and the data were modified accordingly, a Pearson correlation test was run between the values calculated from the GRF curves and the values calculated using the seven different accelerometers methods. When analyzing which method had more success in recreating the force parameters, two aspects of the correlation test were taken into consideration. The first was the number of parameters that correlated to their counterparts calculated from force plate data, the desired result was for all parameters to correlate. The second aspect was the magnitude of the correlation coefficients since high values indicated similar values to force plate calculations which were the desired results. The correlation tests were performed separately for typically developed population alone and the second time for both the typically developed children and children with CP, examining the changes caused by the addition of pathological population.

5.5.1. Self-selected Walking Speeds

When looking at the data of subjects walking at their chosen walking velocities the ANOVA revealed the two side force plates had a main effect on the results. This suggests that at the beginning and/or end of their walks the subjects altered their gait, perhaps as a result of starting or stopping to walk, Interestingly, this effect of force plate for the TD group was only found for one of the plates at the end of the walk (force plate 4 in Figure 2.2).but for children with CP both end force plates (1 and 4 in Figure 2.2) had an effect on the results and therefore both of them were taken out. Only data from the middle two plates were used for subsequent analyses.

For typically developed children (Figure 4.20) it was found that the only method that was able to predict all eight force parameters was the method that included the two accelerometers placed on the back, sacrum and upper back. Peak propulsion and braking force values were the hardest to predict using this method and for the remaining six parameters the calculation accounted for over 80% of the value providing a range of good to excellent correlations. When data of children with CP was added (Figure 4.21), another method was able to predict all values, using the sacrum and corresponding shoulder accelerometers however, it was still the method with two accelerometers on the back that had higher correlation values. Comparison of the correlation values showed some decrease in the correlation coefficient values as well as less of the tightness of the correlation values seen in typically developed children.

The averaged correlation value (Table 4.15) supports the conclusion as well demonstrating that the method consisting of two accelerometers on the back is consistent in producing the best results for both the typically developed population as well as both populations of typically developed children and children with CP.

5.5.2. Results for trials in which subjects walked faster than their chosen walking speed

The ANOVA results for the analysis of the trials in which subjects walked at a faster velocity than their normal walking velocity revealed the same results that were found for the trials in which subjects walked at their chosen walking velocity. The two end force plates were taken out of the analysis and correlation tests were run for the two middle force plates. The correlation results (Figure 4.22) revealed that none of the methods was able to predict all of the parameters. A decrease in the correlation values was seen for all methods affecting also the one parameter that seemed stable for all methods and both populations in the chosen velocity walking condition, the vertical impulse. This might suggest a hardware problem and sensitivity to the increase in movement caused by the faster walking velocity.

When adding the results of children with CP to the correlation test none of the methods could predict the peak propulsion force resulting in the inability of any of the methods to predict all eight parameters. The method consisting of the two back accelerometers produced the best results for the remaining seven parameters with slight improvement in correlation coefficient values suggesting bigger sample sizes would be able to produce more accurate results. The average correlation value (Table 4.15) supported this finding as well demonstrating the ability of this method to produce the most accurate results in both populations.

5.5.3. Results for trials in which subjects walked slower than their chosen walking speed

The ANOVA results for the analysis of the trials in which subjects walked at a slower velocity than their normal walking velocity revealed different results than the ones received for the other two walking conditions. Unlike the previous two conditions, it was revealed that the values for the first and third force plates were different from the values received for the second and fourth force plates. As a result analysis was run on the two sets of force plates separately. These results could be due to the fact that when subjects walk slower than their normal walking speed they had more time to think and adjust their walk causing the changes seen in the ANOVA analysis. Their chosen and faster walking velocities gave less time to think and therefore the walk was more natural. Consistency was seen in the ability of the same method, two accelerometers on the back, to best predict all eight parameters when looking at the first and third force plate values. Comparing the two sets of force plates for typically developed children showed less of a correlation when looking at the values for the second and fourth force plates.

When children with CP were added into the analysis the ability to predict the peak propulsion force diminished but some correlation coefficient values improved suggesting again larger sample sizes can provide less variable results. The correlation for the second and fourth force plates with children with CP added to the analysis produced the weakest results confirming the changes and variability seen in slow walking.

The average correlation value (Table 4.16) demonstrated that overall the strongest method for calculating GRF characteristics was the method that consisted of two accelerometers on the back.

Chapter 6: Conclusions

This work consisted of two parts. The first was to examine the GRFs in typically developed children and children with CP. Examining the GRFs in normal and pathological populations support their use in clinical decision making and gait analysis. The usefulness of GRFs in the examination of gait led to the second body of work which focused on the development of an alternative, simplified and less expensive method of determining these measures. The first step in the process of examination and analysis of GRFs was verifying that the data are reliable, representative, and are not affected by the lab settings. The strength of GRF data relies in the measure being almost completely independent of human interference. Unlike motion analysis, which requires placement of markers on specific bony landmarks and therefore can have inaccuracies due to poor placement and skin movement, as long as the foot strikes the force plate cleanly, reliable GRF data can be collected.

In this work, a GRF consistency test that had been developed and validated in adults (Seliktar, 1979) was applied to pediatric groups to determine consistency of children's gait cycles. There have been no previously reported applications of consistency tests to GRF data in children. The results of the consistency test showed children also demonstrate consistent gait patterns, providing a simple means to identify those gait cycles appropriate for inclusion in clinical gait analysis. Further examination of the GRFs was done for those gait cycles that met the consistency criteria.

As has been demonstrated in Figures 2.3-2.6, the GRF curves have specific characteristics that indirectly represent the body's or COM's movement during the gait cycle. These characteristics, 14 parameters in total, were examined for the GRF in the three planes. Significant differences between typically developed children and the pathological populations were present in peak propulsion force, peak braking force the two peak vertical forces, peak lateral force lateral impulse, braking impulse, propulsion impulse, vertical impulse and the braking to propulsion time. This finding is meaningful because the measures quantified the changes in GRFs and can be used to detect group differences between children with typical development as well as between children with CP with unilateral or bilateral lower limb involvement. Our results also demonstrated that children with hemiplegic CP, who only have one involved side, show compensation for their more involved side by altering the timing, magnitude and impulses for propulsion and braking with changes from TD noted in both involved and uninvolved limbs. These findings further highlight the utility of such GRF parameters that are capable of discriminating the gait deviations seen in children with CP.

This work also extended existing analyses of GRF data by developing several new parameters of clinical relevance. The new parameters included the landing stability ratio, propulsion effort ratio and the performance imbalance index. These parameters quantify differences between the two sides of the body, and were able to provide insight into the amount of deviation from symmetry, between the two sides. Such gait asymmetry is especially important in cases of subjects with only one side involvement such as in children with hemiplegic CP.

To further demonstrate clinical applicability, GRF data were analyzed in six case studies (three in chapter 4 and three in the Appendix B). The case studies examined how different clinical scenarios commonly referred for gait analysis, such as pre/post surgery analysis as well as tracking of children over time and analysis of gait without/with orthotics, are better captured using our analytical approach of GRF data. Changes in each case, differences were identified and quantified using GRF data, providing an enhanced picture of the subject's gait performance in different conditions.

The second part of this work focused on the development of an analytical method that could derive the characteristics of GRF curves in the A-P and vertical directions from direct acceleration data. Several assumptions were made in the development and examination of these different analytical methods. The GRF curves are an algebraic summation of the mass-acceleration products of all the body segments throughout the gait cycle but it is also the reflection of the COM's movement. It was decided to focus on the trunk's movement which holds most of the body's mass as well as the approximated location of the body's COM, the sacrum. An accelerometer was placed on the sacrum and three additional accelerometers were placed on the trunk in order to examine whether additional movement of the trunk could improve the approximation of the COM's location and acceleration. During a gait cycle, the GRF time series consist of single support phases, in which the body is supported on one leg while the other leg swings forward and double support phases where the body moves with both feet on the ground. The COM's acceleration was only used to calculate the forces during the single support phase using a cubic spline function to interpolate the missing data for the double support phases. The acceleration data were calculated using two different filter methods, and five different combinations of the four accelerometers placed on the trunk. In all cases, acceleration data were multiplied by the subject's mass to calculate the force (e.g. F=ma).

The different methods' ability to predict the GRFs was assessed by taking eight characteristics of the GRFs and comparing their values to ones calculated from the force plate data. It was found that the method that most highly correlated in its predicted force values to force plate data was the method that used two accelerometers placed on the back, one accelerometer placed on the sacrum and the other on the T3 region. This method also retained the highest correlation to the force plates GRF parameters with changes in the walking velocity, and for the population of CP children. Changes were noted in the correlation values, some values improved and other decreased, for the CP population. A larger sample of both TD and clinical populations is needed to determine whether these differences generalize to the larger groups. There were also patterns noted as to which GRF characteristics were the hardest to predict from accelerometer data. In particular, the peak propulsion and braking forces were less predictable, and future work should explore whether the use of additional accelerometers on the shanks or ankles improves these parameters. Walking speed also impacted the consistency of the derived GRF. Such change for the faster walking speed could be due to hardware sensitivity, and the use of a different hardware accelerometry system may improve the results. The slow walking speed and its sensitivity to sides, reflect that at a slower walking speed subjects exhibit more variability as they have extra time to consciously think about their walking patterns. The slower walking speeds may not provide the best condition for development of novel techniques, but are important for consideration as many clinical populations use slower walking velocities.

The work done clearly indicates that there is valuable information within the GRF data in children. However there are some limitations when working with GRFs, especially when dealing with children with CP. In this work only mild cases of children with CP who did not use assistive devices were included. GRF data cannot be measured directly using force plates when assistive devices, such as crutches and walkers, are used. When assistive devices are used some of the force exerted by the body is applied to the device and therefore would require force transducers on the assistive devices in order to accurately measure the forces exerted by the body. Further, the assistive devices alter trunk mechanics and would impact the COM typical progression. In order to show the validity of GRFs in gait analysis of clinical populations, a larger sample of children with pathology than just the six cases presented in this work have to be examined.

In summary, the contributions of this work to the field of gait analysis in typically developed children and children with CP include the following:

1. Defined the range of values for which a gait cycle in children can be considered a consistent gait cycle.
2. Identified the GRF characteristics that best capture gait deviations found in children with CP.
    2.1. In children with diplegic CP, the parameters with a significance difference from typically developed children included the peak propulsion force, peak

braking force the two peak vertical forces, peak lateral force lateral impulse and the braking to propulsion time.

   2.2. In children with hemiplegic CP, significant differences were found for the non affected leg for values of the vertical impulse.

   2.3. In the affected leg of children with CP, significance difference were found for the peak propulsion force, second peak of vertical force, braking impulse, propulsion impulse, vertical impulse and the ratio of braking to propulsion time.

   2.4. Correlations of the GRF parameters within the four different populations showed how these parameters relate to one another and the difference in relationships when there are gait deviations.

3. Defined parameters of stability and balance based on the GRF data.

   3.1. The propulsion effort ratio and performance imbalance index were significantly difference in children with hemiplegic CP when compared to the values found in typically developed children.

4. Application of the GRF's analysis techniques to six case studies of children with CP.

5. Developed a method for the derivation of force characteristics from COM's acceleration. The method that was closest in producing the GRF values included two accelerometers placed on the sacrum and upper back.

6. Identified the relationship between EE values, measured using an Actiheart, which showed that in both typically developed children and children with CP, the

GRF parameter that had the best, moderate to good, correlation to EE was the peak braking force.

Chapter 7: Future Work

Future work should apply the derived GRF measures to clinical research applications with a more broadly focused integration of this data with existing gait analysis and clinical measures. Assessing the responsiveness of these measures to pre-post interventions, between group differences, or to detect changes in ambulation due to natural history is needed. Having such quantitative measures to assess changes in gait patterns as well as compare both sides of the body can provide a useful tool for clinicians and doctors. It would be necessary to examine the results and conclusions received in this work on a larger homogenous sample or a larger clinical sample to strengthen the conclusions of this work.

The results for the energy expenditure portion of prospective data collection revealed interesting results which would benefit from further analysis. The first would be to validate the use of the Actiheart to determine energy efficiency during ambulation in children with CP. The second would be to collect energy expenditure and GRF data on a larger sample size to examine whether the results found in this work hold up. Once these steps are taken reliable conclusions on the relationship between energy expenditure and GRF data can be made.

The data and results from the accelerometer measures support the continuation of this line of work as they show GRFs characteristics can be assessed using the acceleration of the body's COM. To further develop the results of this study it would be necessary to increase the sample size as well as examine whether adding accelerometers on the shanks or ankles can help in assessing those GRFS characteristics that were less successfully captured by the trunk's acceleration alone, especially the peak propulsion force. The results of this work should be further explored and validated on a larger sample of typically developed children, with eventual expansion into clinical population such as children with CP. Once the most accurate method of deriving GRF data from acceleration data is determined it should be tested on children with gait abnormalities. Finding the combination of accelerometer data that most accurately characterizes GRF would enable the development of a device that could measure GRF data at lower costs and without being confined to a closed laboratory environment. It would also increase the opportunities in studying GRFs as they could be measured continuously without much difficulty.

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Appendix A: Quantitative results for the GRF analysis

Table A1: Quantitative results of the consistency test for typically developed children and children with diplegic and hemiplegic CP.

Population	*Average Consistency value [%] with Std Dev.*	*95% confidence interval*
Typically Developed Children (N=32)	3.44 ± 2.48	0.87
Children with Diplegic CP (N=32)	3.59 ± 2.47	0.88
Children with Hemiplegic CP (N=42)	4.09 ± 2.58	0.79
Entire population with no regards to condition (N=106)	3.77 ± 2.57	0.15

Table A2: Quantitative results for the calculations of propulsion impulse to braking impulse ratio.

Population	*Propulsion impulse/ Braking impulse with Std Dev.*	*95% confidence interval*
Typically Developed Children (N=54)	1.04 ± 0.20	0.05
Children with Diplegic CP (N=40)	1.03 ± 0.29	0.09
Children with Hemiplegic CP- non affected side (N=44)	1.14 ± 0.33	0.10
Children with Hemiplegic CP- affected side (N=44)	0.98 ± 0.35	0.10

Table A3: The quantitative results for the calculations of braking and propulsion impulses for the four populations.

Population	***Braking Impulse with Std. Dev. (95% confidence interval)***	***Propulsion Impulse with Std. Dev. (95% confidence interval)***
Typically Developed Children (N=54)	-12.02 ± 3.94 (1.05)	12.33 ± 3.94 (1.05)
Children with Diplegic CP (N=40)	-12.00 ± 6.57 (2.04)	12.01 ± 6.48 (2.01)
Children with Hemiplegic CP- non affected side (N=44)	-13.44 ± 5.43 (1.60)	14.37 ± 5.36 (1.58)
Children with Hemiplegic CP- affected side (N=44)	-12.25 ± 4.64 (1.37)	11.25 ± 4.25 (1.26)

Table A4: Results for the calculations of the normalized braking and propulsion impulses for the four populations.

Population	***Normalized braking Impulse with Std. Dev. (95% confidence interval)***	***Normalized propulsion Impulse with Std. Dev. (95% confidence interval)***
Typically Developed Children (N=54)	-0.0277 ± 0.0051 (0.0013)	0.0282 ± 0.0049 (0.0013)
Children with Diplegic CP (N=40)	-0.0276± 0.0081 (0.0025)	0.0277 ± 0.0080 (0.0025)
Children with Hemiplegic CP- non affected side (N=44)	-0.0272 ± 0.0085 (0.0025)	0.0288 ± 0.0060 (0.0018)
Children with Hemiplegic CP- affected side (N=44)	-0.0244 ± 0.0062 (0.0018)	0.0228 ± 0.0067 (0.0023)

Table A5: Quantitative Results of the landing stability and propulsion effort ratios calculations.

Population	***Landing Stability Ratio with Std Dev.***	***95% confidence interval***	***Propulsion Effort Ratio with Std Dev.***	***95% confidence interval.***
Typically Developed Children (N=54)	0.98 ± 0.23	0.06	0.96 ± 0.15	0.04
Children with Diplegic CP (N=40)	0.95 ± 0.42	0.13	0.99 ± 0.24	0.08
Children with Hemiplegic CP (N=44)	0.96± 0.36	0.11	0.80 ± 0.22	0.06

Table A6: Quantitative results of the performance imbalance index calculations.

Population	***Performance imbalance index with Std. Dev.***	***95% confidence interval.***
Typically Developed Children (N=54)	0.96 ± 0.13	0.04
Children with Diplegic CP (N=40)	0.94 ± 0.14	0.04
Children with Hemiplegic CP (N=44)	0.86 ± 0.15	0.04

Appendix B: Additional case studies

Case Study B.1:

This is a male with diplegic CP who was seen for a gait analysis at the ages of 10 and 11. He was seen both times in barefoot condition and visually no significant differences were seen. GRF curves and quantitative results are presented in Figures B.1-B.3 and Table B.1. The results show a decrease in the value of the peak forces in all directions but no change of the existing trends. This could be a result of increased weakness in the muscles as well as increased stiffness that is common in children during puberty. The subject seems to be spending more time in push off as indicated by the braking to propulsion time ratio. The changes due to time, weakness or stiffness of the muscles, is also apparent in propulsion effort ratio and performance imbalance index in which one leg seems to have become more weaker than the other when compared to the previous year.

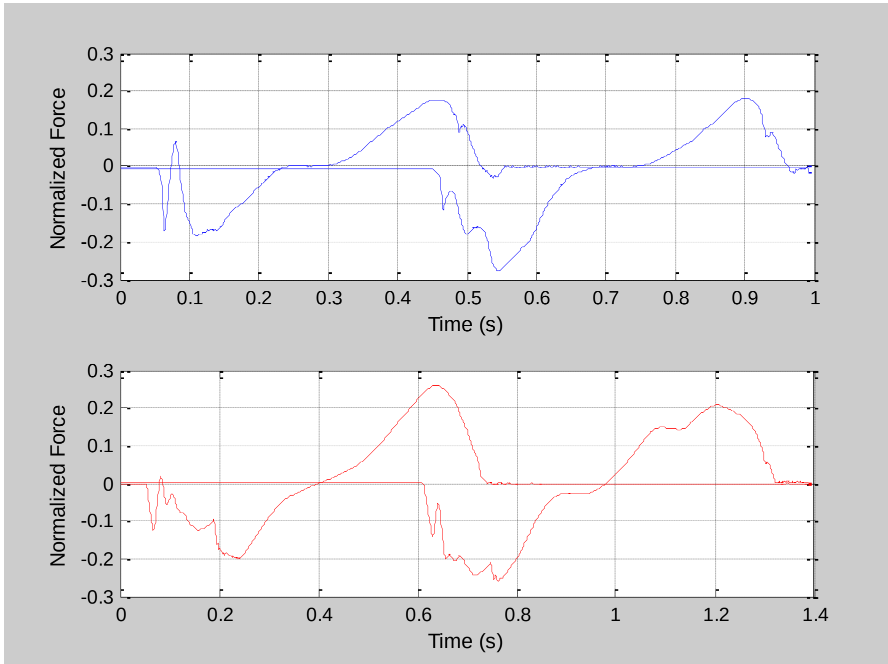

Figure B.1: GRFs in the A-P direction over time, at age 10 (top figure) and age 11 (bottom figure) for case study B.1. The first force in both figures is of the left leg and the second of the right.

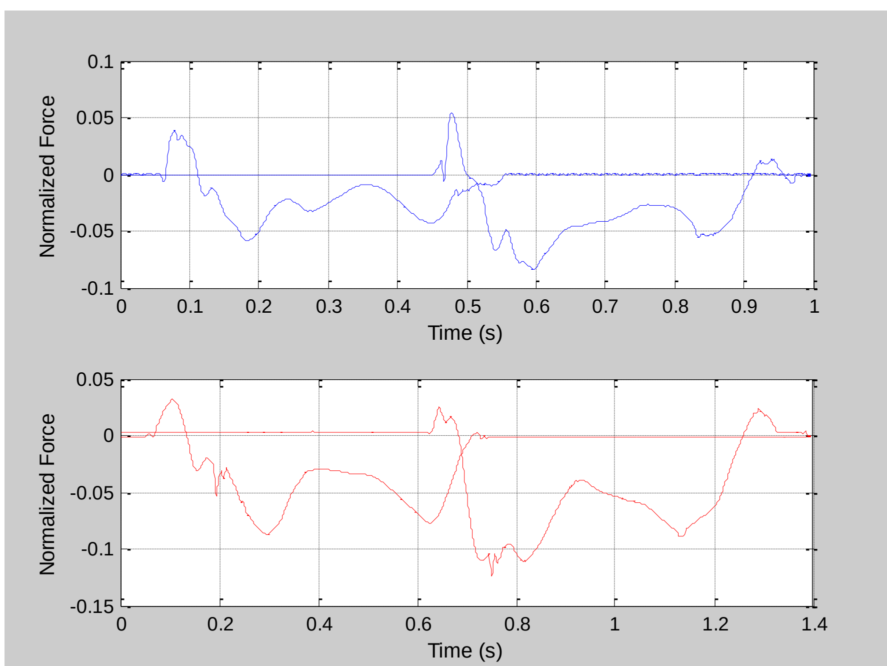

Figure B.2: GRFs in the M-L direction over time, at age 10 (top figure) and age 11 (bottom figure) for case study B.1. The first force in both figures is of the left leg and the second of the right.

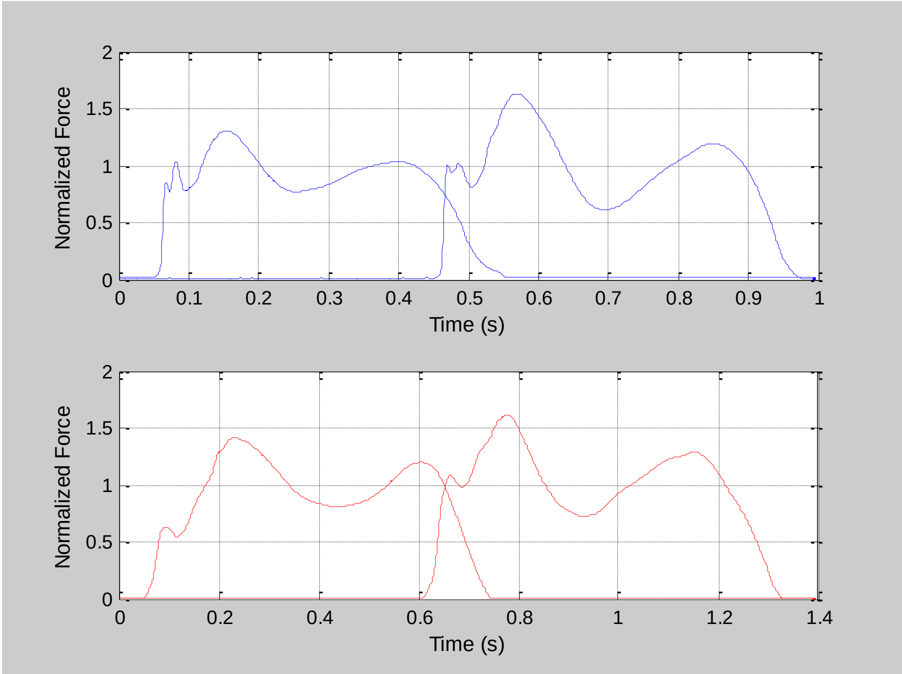

Figure B.3: GRFs in the vertical direction over time, at age 10 (top figure) and age 11 (bottom figure) for case study B.1. The first force in both figures is of the left leg and the second of the right.

Table B.1: Quantitative results for case study B.1

	Barefoot- age 10	***Barefoot- age 11***
Maximum braking force [%BW]	-25.67 ± 5.17	-20.00 ± 2.23
Maximum propulsion force [%BW]	20.20 ± 2.04	14.63 ± 1.65
First Maximum of the vertical force [%BW]	150.31 ± 14.08	135.10 ± 8.24
Second Maximum of the vertical force [%BW]	113.53 ± 11.51	103.08 ± 4.28
Minimum Vertical Force [%BW]	64.17 ± 7.80	65.63 ± 3.60
Maximum Lateral Force [%BW]	8.83 ± 2.19	6.78 ± 1.89

Table B.1: Quantitative results for case study B.1 (continued).

	Barefoot- age 10	***Barefoot- age 11***
Maximum Medial Force [%BW]	3.88 ± 1.38	2.85 ± 1.93
Braking Impulse	-0.030 ± 0.008	-0.022 ± 0.003
Propulsion Impulse	0.027 ± 0.003	0.023 ± 0.003
Vertical Impulse	0.53 ± 0.04	0.50 ± 0.03
Lateral Impulse	0.020 ± 0.006	0.022 ± 0.0016
Medial Impulse	0.0013 ± 0.0004	0.0008 ± 0.0006
Braking/propulsion time	0.72 ± 0.19	0.58 ± 0.12
Landing stability ratio	0.63 ± 0.10	0.74 ± 0.13
Propulsion effort ratio	1.05 ± 0.15	0.80 ± 0.05
Performance imbalance index	1.11 ± 0.33	0.77 ± 0.05

Case Study B.2:

This is a 10 year old female with diplegic CP who was seen for a gait analysis with and without her articulated ankle-foot orthotics. Table B2 presents the analysis of the force parameters in barefoot and braces conditions. The results for the barefoot and braces conditions show no significant changes. There is a decrease in the values of the peak propulsion and braking forces which could be a result of limited mobility in the braces. An improvement in the minimum vertical force between the two peak forces is seen with values increasing towards the ones seen in typically developed children.

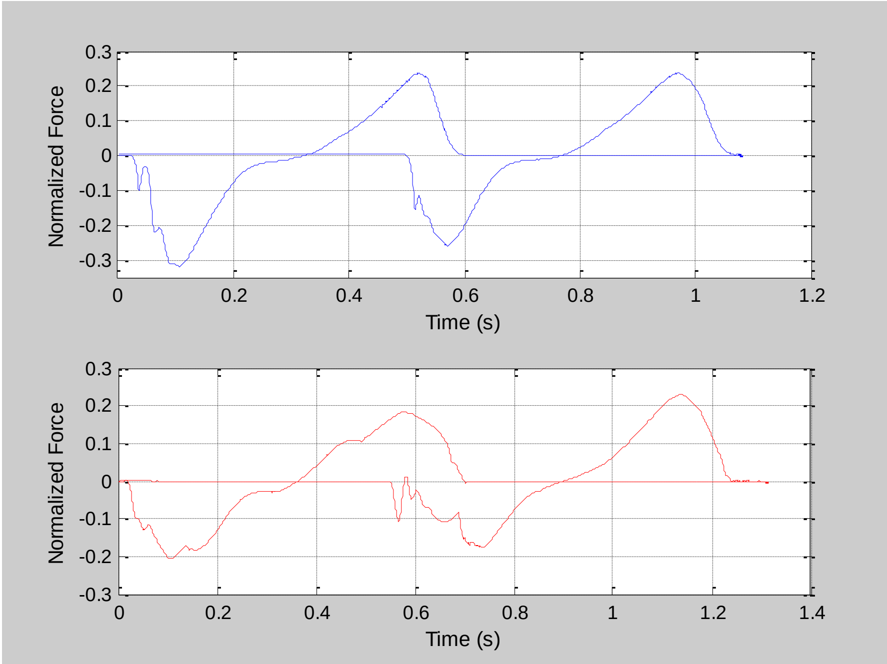

Figure B.4: GRFs in the A-P direction without (top figure) and with (bottom figure) braces for case study B.2. The first force in both figures is of the right leg and the second of the left.

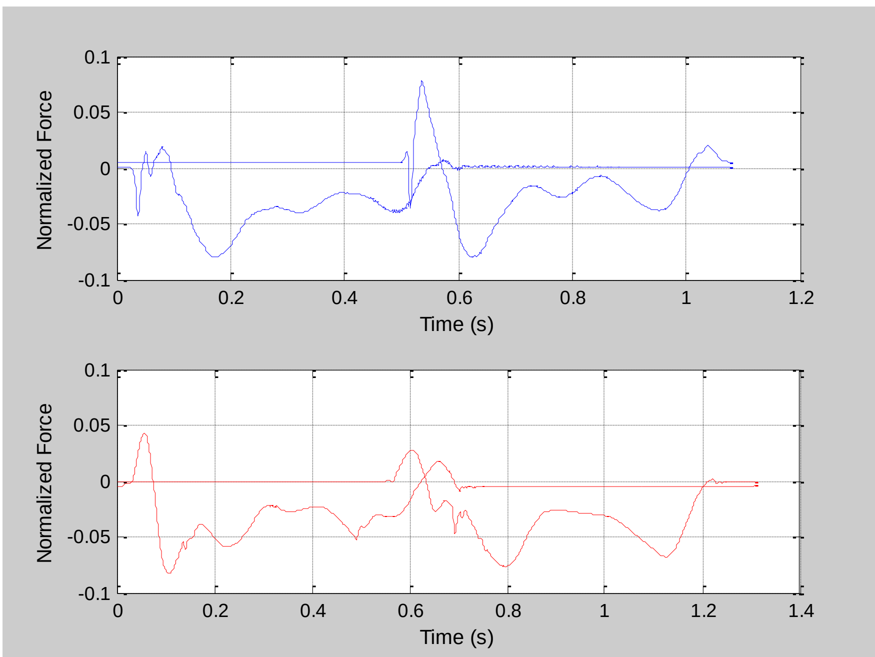

Figure B.5: GRFs in the M-L direction without (top figure) and with (bottom figure) braces for case study B.2. The first force in both figures is of the right leg and the second of the left.

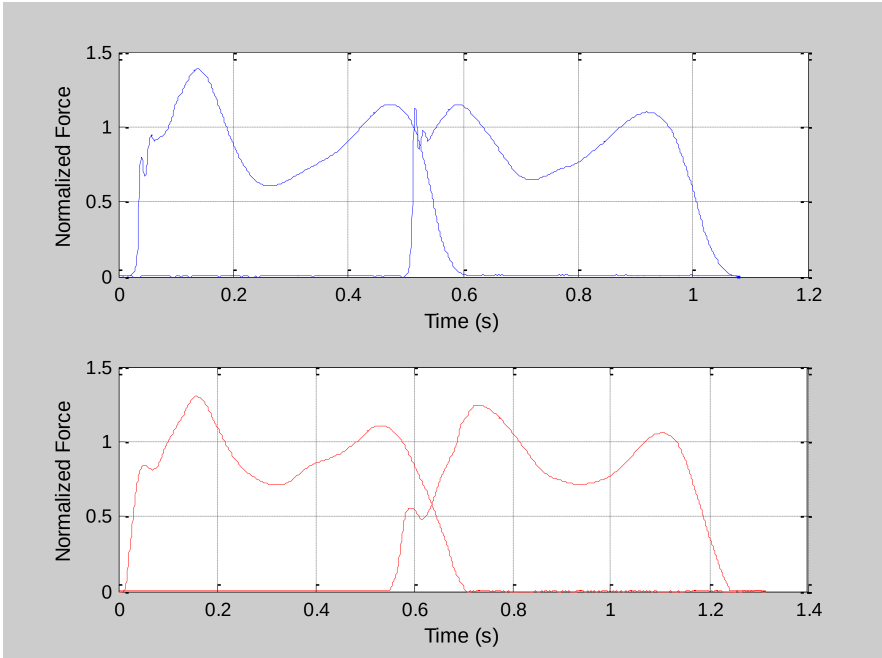

Figure B.6: GRFs in the vertical direction without (top figure) and with (bottom figure) braces for case study B.2. The first force in both figures is of the right leg and the second of the left.

Table B.2: Quantitative results for case study B.2.

	Barefoot	Braces
Maximum braking force [%BW]	-30.21 ± 5.28	-22.34 ± 2.88
Maximum propulsion force [%BW]	25.80 ± 2.46	20.06 ± 2.28
First Maximum of the vertical force [%BW]	130.42 ± 12.29	135.14 ± 8.62
Second Maximum of the vertical force [%BW]	111.80 ± 5.29	109.54 ± 2.75
Minimum Vertical Force [%BW]	59.88 ± 5.25	66.83 ± 4.62
Maximum Lateral Force [%BW]	11.10 ± 2.72	10.21 ± 2.72
Maximum Medial Force [%BW]	1.29 ± 0.78	2.83 ± 1.14

Table B.2: Quantitative results for case study B.2 (continued).

	Barefoot	Braces
Braking Impulse	-0.031 ± 0.006	-0.032 ± 0.005
Propulsion Impulse	0.032 ± 0.003	0.035 ± 0.002
Vertical Impulse	0.50 ± 0.017	0.52 ± 0.024
Lateral Impulse	0.032 ± 0.01	0.027 ± 0.009
Medial Impulse	0.0006 ± 0.0007	0.0009 ± 0.0005
Braking/propulsion time	1.00 ± 0.15	0.99 ± 0.21
Landing stability ratio	1.12 ± 0.16	0.82 ± 0.14
Propulsion effort ratio	1.02 ± 0.24	0.99 ± 0.07
Performance imbalance index	1.06 ± 0.13	1.04 ± 0.16

Case Study B.3:

This is an 8 year old male with left hemiplegic CP who was seen for a gait analysis with and without a left molded ankle-foot orthosis. The comparison seen in Table B2 examines the differences between the affected and non affected sides in barefoot and braced conditions. The brace is worn with shoes therefore in the brace condition the non affected side has only a shoe on.

The subject spends more time in the propulsion phase, especially on the affected side, and the brace does not improve that. Peak barking forces in both conditions are high for the affected leg and suggest control problems during heel strike. Peak forces in the vertical direction show no improvement with the braces on and remain high when compared to typically developed children values. The minimum vertical force value worsens as it decreases farther away from typical values showing the subject needs to flex his knees to be able to push off.

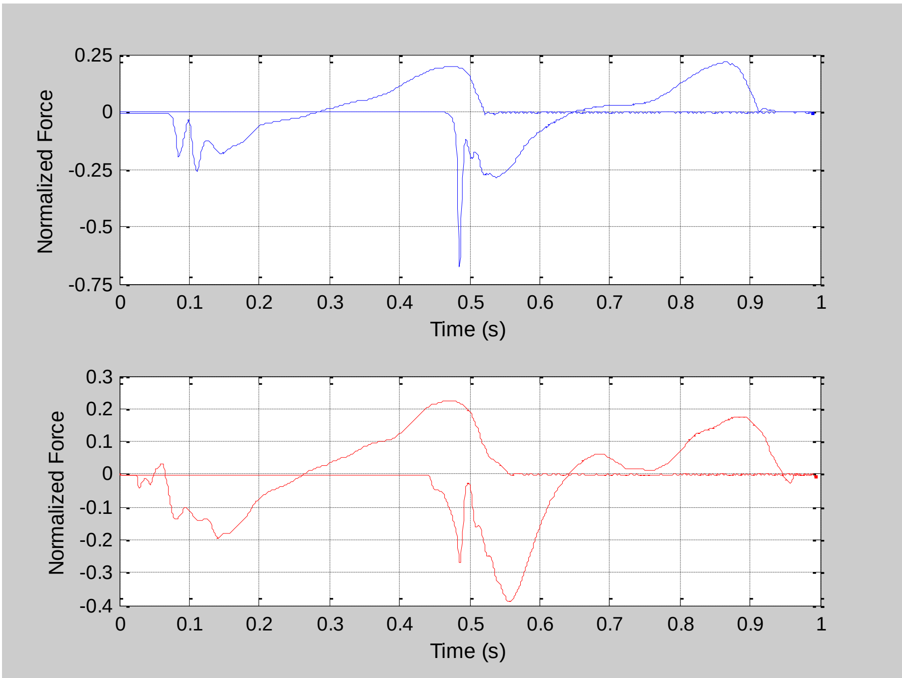

Figure B.7: GRFs in the A-P direction without (top figure) and with (bottom figure) braces for case study B.3. The first force in both figures is of the right leg and the second of the left.

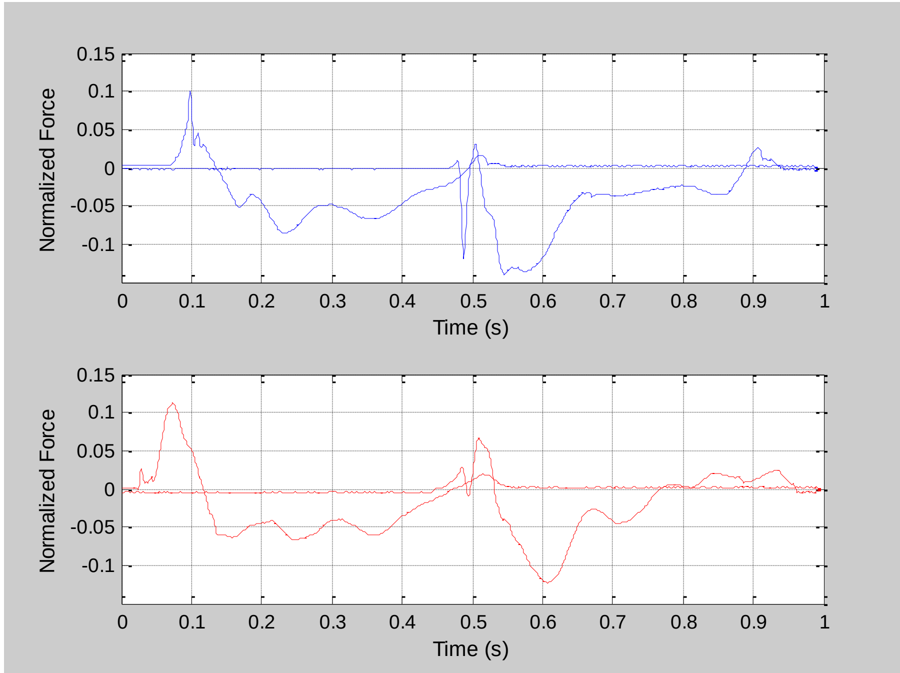

Figure B.8: GRFs in the M-L direction without (top figure) and with (bottom figure) braces for case study B.3. The first force in both figures is of the right leg and the second of the left.

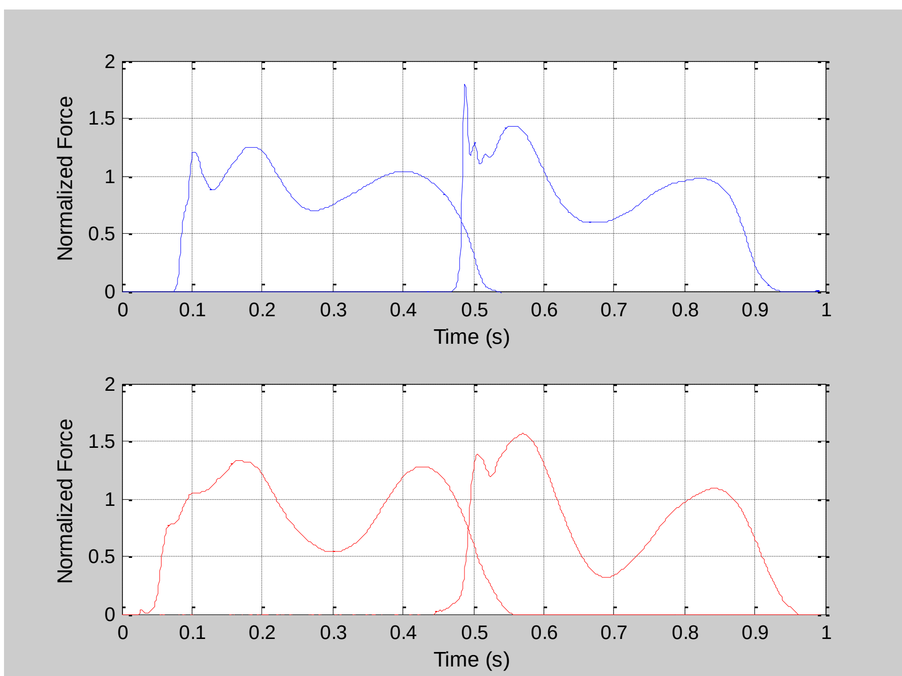

Figure B.9: GRFs in the verical direction without (top figure) and with (bottom figure) braces for case study B.3. The first force in both figures is of the right leg and the second of the left.

Table B.3: Quantitative results for case study B.3.

	Non affected side barefoot	***Affected side barefoot***	***Non affected side with shoe***	***Affected side with brace***
Maximum braking force [%BW]	-25.10 ± 1.73	-32.10 ± 3.13	-24.68 ± 5.95	-32.92 ± 9.20
Maximum propulsion force [%BW]	21.45 ± 1.85	21.77 ± 0.41	25.18 ± 2.60	18.66 ± 0.43
First Maximum of the vertical force [%BW]	130.00 ± 8.00	141.07 ± 1.39	136.64 ± 4.04	150.67 ± 7.50
Second Maximum of the vertical force [%BW]	101.31 ± 2.17	97.46± 7.83	126.37 ± 9.64	104.11 ± 5.00
Minimum Vertical Force [%BW]	75.27 ± 7.69	59.71 ± 0.77	54.40 ± 4.25	41.24 ± 13.54
Maximum Lateral Force [%BW]	10.65 ± 2.41	14.47 ± 0.77	8.53 ± 1.51	7.77 ± 5.77
Maximum Medial Force [%BW]	6.70 ± 2.21	2.73 ± 0.35	10.69 ± 2.18	10.94 ± 5.57
Braking Impulse	-0.030 ± 0.006	-0.036 ± 0.001	-0.031 ± 0.006	-0.042 ± 0.01
Propulsion Impulse	0.032 ± 0.004	0.032 ± 0.002	0.041 ± 0.003	0.034 ± 0.005
Vertical Impulse	0.44 ± 0.03	0.42 ± 0.01	0.50 ± 0.03	0.45 ± 0.02
Lateral Impulse	0.028 ± 0.007	0.027 ± 0.001	0.020 ± 0.004	0.011 ± 0.007
Medial Impulse	0.0025 ± 0.0002	0.0007 ± 0.0004	0.006 ± 0.002	0.011 ± 0.007
Braking/propulsion time	0.91 ± 0.04	0.60 ± 0.01	0.89 ± 0.08	0.56 ± 0.08
Landing stability ratio	1.31 ± 0.34		1.13 ± 0.42	
Propulsion effort ratio	0.98 ± 0.03		1.04 ± 0.37	
Performance imbalance index	1.10 ± 0.15		0.78 ± 0.02	

Appendix C: Quantitative results for the accelerometers analysis

Table C.1. The correlation coefficient values for the Pearson correlation test for typically developed children walking at their self selected walking speed.

	Sacrum Accelerometer	***Sacrum and Upper Back Accelerometers***	***Sacrum and Shoulders Accelerometers***	***Sacrum and Corresponding Shoulder***	***Sacrum and Opposite Shoulder***	***FIR Filter at 3Hz***	***FIR Filter at 4Hz***
Braking Impulse	0.813	0.843	0.765	0.813	0.829	0.816	0.817
Propulsion Impulse	0.850	0.818	0.727	0.656	0.847	0.868	0.879
Vertical Impulse	0.997	0.997	0.997	0.997	0.997	0.998	0.997
Peak Propulsion Force	--	0.422	0.414	--	0.456	--	0.407
Peak Braking Force	0.518	0.641	--	--	--	--	--
1ST Vertical Peak Force	0.816	0.792	0.802	0.805	0.761	0.598	0.686
2nd Vertical Peak Force	0.793	0.822	0.797	0.759	0.806	0.691	0.773
Minimum Vertical Force between two Peaks	0.789	0.863	0.886	0.860	0.872	0.744	0.776

Table C.1-C.8 notes:

1. The results are for the Pearson correlation test between the force characteristics values calculated using the 7 different methods using acceleration and the values calculated from the GRFs data.
2. Empty cells represent cases in which no correlation existed.

Table C.2. The correlation coefficient values for the Pearson correlation test for typically developed children and children with CP walking at their self selected walking speed.

	Sacrum Accelerometer	***Sacrum and Upper Back Accelerometers***	***Sacrum and Shoulders Accelerometers***	***Sacrum and Corresponding Shoulder***	***Sacrum and Opposite Shoulder***	***FIR Filter at 3Hz***	***FIR Filter at 4Hz***
Braking Impulse	0.686	0.728	0.630	0.597	0.748	0.627	0.650
Propulsion Impulse	0.821	0.805	0.805	0.734	0.845	0.824	0.828
Vertical Impulse	0.997	0.997	0.997	0.998	0.996	0.997	0.997
Peak Propulsion Force	--	0.356	--	0.369	--	--	--
Peak Braking Force	0.435	0.662	--	0.389	--	--	
1ST Vertical Peak Force	0.865	0.875	0.749	0.814	0.617	0.774	0.819
2nd Vertical Peak Force	0.452	0.703	0.631	0.557	0.635		0.368
Minimum Vertical Force between two Peaks	0.804	0.898	0.741	0.759	0.703	0.764	0.803

Table C.3. The correlation coefficient values for the Pearson correlation test for typically developed children walking faster than their self selected walking speed.

	Sacrum Accelerometer	***Sacrum and Upper Back Accelerometers***	***Sacrum and Shoulders Accelerometers***	***Sacrum and Corresponding Shoulder***	***Sacrum and Opposite Shoulder***	***FIR Filter at 3Hz***	***FIR Filter at 4Hz***
Braking Impulse	0.667	0.728	--	0.445	0.568	0.813	0.764
Propulsion Impulse	0.434	0.440	0.442	0.436	0.657	--	0.430
Vertical Impulse	0.955	0.928	0.980	0.978	0.992	0.956	0.956
Peak Propulsion Force	--	--	0.544	--	0.569	--	--
Peak Braking Force	--	0.606	--	--	--	--	--
1ST Vertical Peak Force	0.761	0.825	0.814	0.798	0.774	0.587	0.692
2nd Vertical Peak Force	0.869	0.866	0.849	0.843	0.861	0.655	0.766
Minimum Vertical Force between two Peaks	0.779	0.830	0.754	0.694	0.820	--	0.638

Table C.4. The correlation coefficient values for the Pearson correlation test for typically developed children and children with CP walking faster than their self selected walking speed.

	Sacrum Accelerometer	***Sacrum and Upper Back Accelerometers***	***Sacrum and Shoulders Accelerometers***	***Sacrum and Corresponding Shoulder***	***Sacrum and Opposite Shoulder***	***FIR Filter at 3Hz***	***FIR Filter at 4Hz***
Braking Impulse	0.564	0.612	0.437	--	0.510	0.541	0.563
Propulsion Impulse	0.646	0.662	--	0.395	0.734	0.646	0.654
Vertical Impulse	0.970	0.955	0.984	0.983	0.979	0.967	0.968
Peak Propulsion Force	--	--	--	--	--	--	--
Peak Braking Force	0.481	0.663	--	--	--	0.502	0.482
1ST Vertical Peak Force	0.778	0.811	0.676	0.738	0.596	0.572	0.674
2nd Vertical Peak Force	0.757	0.837	0.830	0.843	0.795	0.526	0.635
Minimum Vertical Force between two Peaks	0.766	0.842	0.644	0.623	0.629	0.530	0.759

Table C.5. The correlation coefficient values for the Pearson correlation test for typically developed children walking slower than their self selected walking speed on the first and third force plates.

	Sacrum Accelerometer	***Sacrum and Upper Back Accelerometers***	***Sacrum and Shoulders Accelerometers***	***Sacrum and Corresponding Shoulder***	***Sacrum and Opposite Shoulder***	***FIR Filter at 3Hz***	***FIR Filter at 4Hz***
Braking Impulse	0.821	0.801	0.666	0.702	0.742	0.782	0.809
Propulsion Impulse	0.793	0.829	0.736	0.790	0.758	0.780	0.789
Vertical Impulse	0.998	0.998	0.998	0.998	0.998	0.998	0.996
Peak Propulsion Force	0.385	0.492	0.441	0.568	--	--	--
Peak Braking Force	0.587	0.643	--	0.450	--	0.463	0.535
1ST Vertical Peak Force	0.730	0.703	0.702	0.772	0.633	--	--
2nd Vertical Peak Force	0.546	0.678	0.661	0.479	0.728	0.729	0.700
Minimum Vertical Force between two Peaks	0.541	0.578	0.634	0.498	0.660	--	--

Table C.6. The correlation coefficient values for the Pearson correlation test for typically developed children and children with CP walking slower than their self selected walking speed on the first and third force plates.

	Sacrum Accelerometer	***Sacrum and Upper Back Accelerometers***	***Sacrum and Shoulders Accelerometers***	***Sacrum and Corresponding Shoulder***	***Sacrum and Opposite Shoulder***	***FIR Filter at 3Hz***	***FIR Filter at 4Hz***
Braking Impulse	0.723	0.735	0.661	0.686	0.681	0.703	0.719
Propulsion Impulse	0.734	0.839	0.820	0.776	0.803	0.755	0.750
Vertical Impulse	0.996	0.996	0.996	0.996	0.996	0.997	0.995
Peak Propulsion Force	--	--	--	0.373	--	--	--
Peak Braking Force	0.411	0.617	0.364	--	--	--	0.365
1ST Vertical Peak Force	0.908	0.889	0.739	0.906	--	0.744	0.846
2nd Vertical Peak Force		0.536	0.415	0.364	0.388	--	--
Minimum Vertical Force between two Peaks	0.877	0.910	0.689	0.868	--	0.742	0.800

Table C.7. The correlation coefficient values for the Pearson correlation test for typically developed children and walking slower than their self selected walking speed on the second and fourth force plates.

	Sacrum Accelerometer	***Sacrum and Upper Back Accelerometers***	***Sacrum and Shoulders Accelerometers***	***Sacrum and Corresponding Shoulder***	***Sacrum and Opposite Shoulder***	***FIR Filter at 3Hz***	***FIR Filter at 4Hz***
Braking Impulse	0.862	0.756	0.625	0.737	0.690	0.835	0.862
Propulsion Impulse	0.838	0.755	0.746	0.742	0.810	0.811	0.848
Vertical Impulse	0.997	0.997	0.997	0.998	0.997	0.997	0.998
Peak Propulsion Force	--	--	--	--	--	--	--
Peak Braking Force	0.698	0.664	--	--	0.461	0.617	0.661
1ST Vertical Peak Force	0.830	0.907	0.891	0.905	0.865	0.807	0.827
2nd Vertical Peak Force	0.751	0.829	0.742	0.774	0.706	0.597	0.664
Minimum Vertical Force between two Peaks	0.451	0.537	0.527	0.487	0.517	0.735	0.655

Table C.8. The correlation coefficient values for the Pearson correlation test for typically developed children and children with CP walking slower than their self selected walking speed on the second and fourth force plates.

	Sacrum Accelerometer	***Sacrum and Upper Back Accelerometers***	***Sacrum and Shoulders Accelerometers***	***Sacrum and Corresponding Shoulder***	***Sacrum and Opposite Shoulder***	***FIR Filter at 3Hz***	***FIR Filter at 4Hz***
Braking Impulse	0.810	0.740	0.619	0.709	0.607	0.779	0.807
Propulsion Impulse	0.834	0.742	0.765	0.760	0.780	0.799	0.866
Vertical Impulse	0.996	0.996	0.996	0.997	0.992	0.996	0.991
Peak Propulsion Force	--	--	--	--	--	--	--
Peak Braking Force	0.581	0.676	--	0.442	--	0.509	0.555
1ST Vertical Peak Force	0.787	0.892	0.768	0.478	0.844	0.910	0.918
2nd Vertical Peak Force	0.513	0.743	0.683	0.589	0.656	0.383	0.477
Minimum Vertical Force between two Peaks	0.815	0.808	0.378	--	0.832	0.652	0.730